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This is a request for filing a PROVISIONAL APPLICATION FOR PATENT under 37 CFR 1.53(c).

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TITLE OF THE INVENTION (280 characters max)

HEAD COIL AND NEUROVASCULAR ARRAY FOR PARALLEL IMAGING CAPABLE MAGNETIC RESONANCE SYSTEMS

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Respectfully submitted,

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Date 2/22/2004

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Invention: **HEAD COIL AND NEUROVASCULAR ARRAY FOR PARALLEL IMAGING CAPABLE MAGNETIC RESONANCE SYSTEMS**

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HEAD COIL AND NEUROVASCULAR ARRAY FOR PARALLEL IMAGING CAPABLE MAGNETIC RESONANCE SYSTEMS

FIELD OF THE INVENTION

[01] The invention generally relates to the field of magnetic resonance (MR) imaging and spectroscopy systems and to the various types of local coils used with such systems. More particularly, the invention pertains to local coils, such as those used to image the human head, which are designed to be used with 8-channel MR systems capable of acquiring images using parallel imaging techniques.

BRIEF DESCRIPTION OF RELATED ART

[02] The following information is provided to assist the reader to understand the environment in which the invention disclosed herein will typically be used.

[03] Magnetic resonance imaging (MRI) is a noninvasive method of producing high quality images of the interior of the human body. It allows medical personnel to see inside the body (e.g., organs, muscles, nerves, bones, and other structures) without surgery or the use of potentially harmful ionizing radiation such as X-rays. The images are of such high resolution that disease and other pathological conditions can often be visually distinguished from healthy tissue. Magnetic resonance (MR) systems and techniques have also been developed for performing spectroscopic analyses by which the chemical content of tissue or other material can be ascertained.

[04] MR imaging and spectroscopic procedures are performed in what is known as an MR suite. As shown in Figure 1A, an MR suite typically has three rooms: a scanner room 1, a control room 2, and an equipment room 3. The scanner room 1 houses the MR scanner 10 into which a patient is moved via a slideable table 11 to undergo a scanning procedure, and the control room 2 contains a computer console 20 from which the operator controls the overall operation of the MR system. In addition to a door 4, a window 5 is typically set in the wall separating the scanner and control rooms to allow the operator to observe the patient during such procedures. The

equipment room 3 contains the various subsystems necessary to operate the MR system. The equipment includes a power gradient controller 31, a radio frequency (RF) assembly 32, a spectrometer 33, and a cooling subsystem 34 with which to avoid the build up of heat which, if left unaddressed, could otherwise interfere with the overall performance of the MR system. These subsystems are typically housed in separate cabinets, and are supplied electricity through a power distribution panel 12 as are the scanner 10 and the slideable patient table 11.

[05] An MR system obtains such detailed images and spectroscopic results by taking advantage of the intrinsic properties of the hydrogen atom, which is found in abundance in all cells within the body. The nuclei in hydrogen atoms naturally spin like a top, or precess, randomly in every direction. Furthermore, because they act as tiny dipole magnets, the hydrogen nuclei or "spins" tend to line up in the direction of the magnetic field to which they are exposed. During an MR scan, the entire body or, optionally, any desired region thereof is exposed to such a magnetic field. This compels the nuclei/spins of the exposed region(s) to line up --and collectively form an average vector of magnetization-- in the direction of the magnetic field.

[06] As shown in Figures 1B and 1C, the scanner 10 is comprised of a main magnet 101, three gradient coils 103a-c, and, usually, an RF antenna 104 (often referred to as the whole body coil). Superconducting in nature, the main magnet 101 is typically cylindrical in shape. Within its cylindrical bore, the main magnet 101 generates a strong magnetic field, often referred to as the B_0 or main magnetic field, which is both uniform and static (non-varying). For a scanning procedure to be performed, the patient must be moved into this cylindrical bore, typically while supine on table 11, as best shown in Figures 1B and 1C. The main magnetic field is oriented along the longitudinal axis of the bore, referred to as the z direction, which compels the magnetization vectors of the hydrogen nuclei in the body to align themselves in that direction. In this alignment, the hydrogen nuclei/spins are prepared to receive RF energy of the appropriate frequency from RF coil 104. This frequency is known as the Larmor frequency and is governed by the equation $\omega = \Upsilon B_0$, where ω is the

Larmor frequency (at which the hydrogen atoms precess), Υ is the gyromagnetic constant, and B_0 is the strength of the main magnetic field.

[07] The RF coil 104 is typically used both to transmit pulses of RF energy and to receive the resulting magnetic resonance (MR) signals induced thereby in the hydrogen nuclei. Specifically, during its transmit cycle, the coil 104 broadcasts RF energy into the cylindrical bore. This RF energy creates a radio frequency magnetic field, also known as the RF B_1 field, whose field lines point in a direction perpendicular to the magnetization vectors of the hydrogen nuclei. The RF pulse (or B_1 field) causes the spin-axes of the hydrogen nuclei to tilt with respect to the main (B_0) magnetic field, thus causing the net magnetization vectors to deviate from the z direction by a certain angle. The RF pulse, however, will affect only those hydrogen nuclei/spins that are precessing about their axes at the frequency of the RF pulse. In other words, only the nuclei that "resonate" at that frequency will be affected, and such resonance is achieved in conjunction with the operation of the three gradient coils 103a-c.

[08] Each of the three gradient coils is used to vary the main (B_0) magnetic field linearly along one of the three directions (x,y,z) within the cylindrical bore. Positioned inside the main magnet as shown in Figure 1C, the gradient coils 103a-c when turned on and off very rapidly in a specific manner are able to alter the B_0 field on a very local level and, in doing so, enable the MR system to acquire as quickly as possible the spatial information discussed below. Thus, in conjunction with the main magnet 101, the gradient coils 103a-c can be operated according to various imaging techniques so that the hydrogen nuclei/spins --at any given point or in any given strip, slice or unit of volume-- will be able to achieve resonance when an RF pulse of the appropriate frequency is applied. In response to the RF pulse, the precessing hydrogen nuclei in the selected region absorb the RF energy being transmitted from RF coil 104, thus forcing the magnetization vectors thereof to tilt away from the direction of the main (B_0) magnetic field. When the RF coil 104 is turned off, the hydrogen nuclei/spins begin to release the RF energy they just absorbed in the form of magnetic resonance (MR) signals, as explained further below.

[09] One well known technique that can be used to obtain images is referred to as the spin echo imaging technique. Operating according to this MR sequence, the MR system first activates one gradient coil 103a to set up a magnetic field gradient along the z-axis. This is called the "slice select gradient," and it is set up when the RF pulse is applied and shut off when the RF pulse is turned off. It allows resonance to occur only within those hydrogen nuclei located within a slice of the region being imaged. No resonance will occur in any tissue located on either side of the plane of interest. Immediately after the RF pulse ceases, all of the nuclei/spins in the activated slice are "in phase," i.e., their magnetization vectors all point in the same direction. Left to their own devices, the net magnetization vectors of all the hydrogen nuclei in the slice would relax and thus realign with the z direction. Instead, however, the second gradient coil 103b is briefly activated to create a magnetic field gradient along the y-axis. This is called the "phase encoding gradient." It causes the magnetization vectors of the nuclei within the slice to point, as one moves between the weakest and strongest ends of this gradient, in increasingly different directions. Next, after the RF pulse, the slice select gradient, and the phase encoding gradient have been turned off, the third gradient coil 103c is briefly activated to create a gradient along the x-axis. This is called the "frequency encoding gradient" or "read out gradient," as it is only applied when the MR signal is ultimately measured. It causes the relaxing magnetization vectors to be differentially re-excited, so that the nuclei/spins near the low end of that gradient begin to precess at a faster rate, and those at the high end pick up even more speed. When these nuclei then relax, the fastest ones (those which were at the high end of the gradient) will emit the highest frequency of radio waves and the slowest ones will emit the lowest frequencies. In this manner, the gradient coils 103a-c spatially encode the radio waves to be emitted by the hydrogen nuclei/spins, with each tiny part of the region being imaged being uniquely defined by the frequency and phase of its resonance signal.

[10] The hydrogen nuclei/spins thus collectively emit their radio waves in a specific band of frequencies determined by the B_0 magnetic field of main magnet 101 and the specific spatial variations in the B_0 field made possible by gradient coils 103a-c. During its receive cycle, RF coil 104 detects these miniature radio emissions, which

are often collectively referred to as the MR signal(s). These unique resonance signals are then conveyed to the receivers of the MR system, wherein they are stored as a line of information in a data matrix known as the k-space matrix. The full matrix is built up by successive cycles of conditioning the hydrogen nuclei, perturbing them, and collecting the resultant RF emissions. Using a technique known as Fourier transformation, the MR system then converts the frequency information present in these RF emissions to spatial information representing the distribution of nuclei/spins in the tissue being scanned. Having determined the exact location of the nuclei/spins in space, the MR system can then display a two- or even a three-dimensional image of the body, or region thereof, that was scanned.

[11] When more detailed images of a specific part of the body are needed, a local coil is often used in addition to, or instead of, the whole body coil 104. A local coil can take the form of a volume coil or a surface coil. A volume coil is used to surround or enclose a volume (e.g., a head, an arm, a wrist, a knee or other region) to be imaged. Some volume coils (e.g., for imaging the head and/or extremities) are often referred to as birdcage coils due to their shape. A surface coil, however, is merely fitted or otherwise placed against a surface (e.g., a shoulder, a breast, etc.) of the patient so that the underlying region can be imaged. A local coil can also be designed to operate either as a receive-only coil or a transmit/receive (T/R) coil. A receive-only coil is only capable of detecting the MR signals emitted by the body. A T/R coil, however, is capable of both receiving the MR signals as well as transmitting the RF pulses that produce the RF B_1 magnetic field, which is the prerequisite for inducing resonance in the tissues of the region of interest.

[12] It is well known in the field of MR to use a single local coil, whether surface or volume, to detect the MR signals. According to the single coil approach, a relatively large local coil is used to cover or enclose the entire field of view. Early receiving coils were just linear coils, meaning that they could detect only one of the two (i.e., vertical and horizontal) quadrature components of the MR signals produced by the region of interest. Subsequent receiving coils, however, employed quadrature mode detection, meaning that they could intercept both the vertical and horizontal

components. Compared to linear receiving coils, quadrature receiving coils enabled MR systems to provide images for which the SNR was much improved, theoretically by as much as 41%. Even with the improvement brought with quadrature mode detection, the single coil approach still provided images whose quality invited improvement. The disadvantage of the single coil approach is attributable to just one coil structure being used to acquire the MR signals over the entire field of view.

[13] With the advent of MR systems equipped with multiple receivers, phased array coils were developed to overcome the shortcomings with the single coil approach. Instead of one large local coil, the phased array approach uses a plurality of smaller local coils (also referred to as "coil elements"), with each such coil element covering or enclosing only a portion of the field of view and its output typically routed to one receiver or channel of the MR system. In a phased array coil system having two such coil elements, for example, each element would cover or enclose approximately half of the field of view, with the two coil elements being partially overlapped for purposes of magnetic isolation. The two coil elements would acquire the MR signals from their respective portions simultaneously, and they would not interact adversely due to the overlap. Because each coil element covers only half of the field of view, each such coil element is able to receive the MR signals at a higher SNR ratio for that portion of the field of view within its coverage area. The smaller coil elements of the phased array thus collectively provide the MR system with the signal data necessary to generate an image of the entire field of view that is higher in resolution than what can be obtained from a single large local coil.

[14] One example of a phased array coil is the neurovascular array (NVA). An NVA is typically comprised of a series of local coils that are used to image the head, neck and cervical spine regions of the body. A particular NVA and the head coil used therewith are disclosed in U.S. Patent 6,356,081 to *Misic* and U.S. Patent 6,344,745 to *Reisker et al.*, respectively, which are incorporated herein by reference. Such head coils, or birdcages, typically have a pair of circular end rings which are bridged by a plurality of equispaced straight rods. The birdcage coil disclosed in the '745 patent has end rings of different size, however, and the rods that extend therebetween are

tapered accordingly. This improves the homogeneity of the magnetic flux density throughout the head coil, particularly in its XZ and YZ imaging planes.

[15] As noted above, the spatial information acquired through magnetic resonance techniques is encoded through the application of rapidly-switched magnetic field gradients and RF pulses. The speed of an MR scanning procedure is therefore determined by how quickly this spatial encoding may be performed. Most of the fast imaging sequences now in use (EPI, FLASH, TSE, or BURST, for example) achieve their high speeds by optimizing the switching rates and patterns of the gradients and RF pulses. Nevertheless, the one feature common to these techniques is that they all acquire data in a sequential fashion. Whether the required data set (i.e., the k-space data matrix) is filled in a rectangular raster pattern, a spiral pattern, a rapid series of line scans, or some other novel order, it is still acquired one point and one line at a time, with each separate point or line of data requiring a separate application of field gradients and/or RF pulses. The speed at which MR images can be obtained using these techniques is thus limited because they acquire data sequentially in time. Further shortcomings with sequential data acquisition techniques involve potential adverse effects on patient safety.

[16] SMASH, which stands for "SiMultaneous Acquisition of Spatial Harmonics," is a partially parallel imaging technique, which exploits the geometry of an RF coil array to encode multiple lines of MR image data simultaneously, thereby multiplying the speed of existing sequential imaging sequences by an integer factor. In experiments using commercially available coil arrays, SMASH has been used to accelerate a number of fast imaging sequences, without increasing gradient switching rates or RF power deposition.

[17] Nearly all existing sequential rapid imaging sequences may be accelerated in this manner, and, to date, SMASH has been successfully tested with a wide range of sequence types, including TSE, RARE, HASTE, TFE, FLASH, TrueFISP, EPI, and BURST. Both two-dimensional and three-dimensional acquisitions are amenable to acceleration using SMASH. Whereas the ultimate speeds of most sequential imaging techniques are limited by physical and physiologic constraints on gradient switching

rate and RF power deposition, achievable SMASH imaging speeds are limited in principle only by the number and arrangement of RF array elements which may reasonably be constructed and interfaced with an MR scanner. The improvements in imaging efficiency afforded by SMASH may be put to use in a number of ways, including: (1) reduction in breath-hold times for clinical MR scans, to increase patient compliance and comfort; (2) reduction in the overall duration of longer scans, once again increasing comfort and compliance, and also increasing the throughput of clinical MR scanners and the cost-effectiveness of MR diagnosis; (3) improvements in temporal resolution (i.e., shorter image acquisition intervals), minimizing undesired effects of physiologic motion while allowing accurate tracking of time-dependent phenomena; (4) improvements in the spatial resolution which may be achieved in a given imaging time; and (5) improvements in image quality resulting from a reduction in time-dependent artifacts (due to motion, susceptibility, relaxation, etc.).

[18] Sensitivity encoding (SENSE) is another parallel imaging technique. It can be used to reduce scan time in MRI considerably. The spatial information related to the coils of a receiver array are utilized for reducing conventional Fourier encoding. SENSE can, in principle, be applied to any imaging sequence and k-space trajectories.

[19] ASSET (Array Spatial Sensitivity Encoding Technique) is yet another parallel imaging technique. Developed by General Electric Medical Systems (GEMS), ASSET uses the unique geometry of phased array coils to spatially encode the image faster. The ASSET technique can be used to scan faster, improve spatial resolution and/or increase coverage.

[20] One MR system that is capable of acquiring images of a region of interest using parallel imaging techniques is the GEMS Signa® 8-channel 1.5 Tesla MR system. Due to the development of such MR systems, there is now a need to provide local coils and the associated interface circuitry to take advantage of the faster parallel imaging capabilities offered by such new MR systems.

[21] The prior art head coils disclosed in the above-cited patents were not originally built for operation with MR systems capable of acquiring images of the head using parallel imaging techniques. The head coil disclosed in the '745 patent acquires its images by means of overlapping signal patterns, i.e., the two quadrature modes overlap and are acquired simultaneously. Because of that overlap, the head coil is not appropriate for acquiring images using parallel imaging techniques. It would therefore be quite advantageous to develop a head coil that is capable not only of providing the homogeneity of the prior art head coil disclosed in the '745 patent but also of acquiring images using parallel imaging techniques.

BRIEF DESCRIPTION OF THE DRAWINGS

[22] The invention, and particularly its presently preferred and alternative embodiments and related aspects, will be better understood by reference to the detailed disclosure below and to the accompanying drawings, in which:

[23] **Figure 1A** illustrates the layout of an MR suite inclusive of the scanner room in which the scanner and patient table are located, the control room in which the computer console for controlling the scanner is situated, and the equipment room in which various control subsystems for the scanner are sited.

[24] **Figure 1B** shows a scanner and patient table of the type shown schematically in Figure 1A.

[25] **Figure 1C** is a more detailed view of the MR system shown in Figures 1A and 1B showing the computer console and the various subsystems located in the control and equipment rooms and a cross-section of the scanner and patient table situated in the scanner room.

[26] **Figure 2** is a schematic circuit diagram for a presently preferred embodiment of a receive-only tapered head coil capable of being used as part of a neurovascular array.

[27] **Figures 2A-2G** also illustrate the presently preferred embodiment of the tapered head coil of Figure 2, with minor variations in componentry.

[28] **Figure 3** is a generalized diagram of the tapered head coil of Figure 2 showing, from the perspective of its superior end, the eight electrically-adjacent primary resonant substructures of the head coil and their angular distribution and alternating arrangement, small adjacent to large, about the birdcage-type structure.

[29] **Figure 4** is a schematic circuit diagram for a preferred embodiment of an anterior neck coil capable of being used as part of a neurovascular array.

[30] **Figure 5** is a schematic circuit diagram for a preferred embodiment of a posterior cervical spine coil capable of being used as part of a neurovascular array.

[31] **Figures 6A-6B and 7A-7B** illustrate a schematic circuit diagram of a presently preferred embodiment of an interface circuit for coupling/multiplexing the coil elements of the head coil and the anterior neck and posterior C-spine coils of a neurovascular array to a multi-channel MR system.

[32] **Figure 8** is a generalized diagram of a head coil according to a second embodiment of the invention showing, from the perspective of its superior end, six electrically-adjacent primary resonant substructures and their even angular distribution about the birdcage-type structure.

[33] **Figure 9** is a generalized diagram of a head coil according to a third embodiment of the invention showing, from the perspective of its superior end, twelve electrically-adjacent primary resonant substructures and their even angular distribution about the birdcage-type structure.

[34] **Figure 10** is an isometric view of a preferred embodiment of a neurovascular array housing, inclusive of the base section, the head coil section, the anterior neck coil section, and the posterior cervical spine coil section.

[35] **Figures 11** and **12** are exploded views of the neurovascular array housing of Figure 10, showing (i) the cover and outer and inner subhousings of the head coil section; (ii) the mirror assembly for the head coil section, (iii) the outer and inner subhousings of the anterior neck section, (iv) the paddle arms and pivot damper

assemblies for the anterior neck section, (v) the cervical Spine section and (vi) the base and cover for the base section.

[36] **Figure 13** is a left side view of the neurovascular array housing, which illustrates (i) the head coil section at the inferiormost end of its travel (i.e., closed position) and (ii) the pivotable anterior neck section lowered to a fully engaged position.

[37] **Figure 14** is a left side view of the neurovascular array housing, which shows (i) the slidable head coil section having been moved to the superiormost end (i.e., open position) and (ii) the pivotable anterior neck section in an upright position.

[38] **Figure 15** is a left side view of the neurovascular array housing, which shows (i) the slidable head coil section in the open position and (ii) the pivotable anterior neck section lowered to the fully engaged position.

[39] **Figure 16** is an exploded view of the head coil section showing the cover thereof, the inner and outer subhousings thereof and the sliders thereof.

[40] **Figure 17** is a left side view of the head coil section showing the roller assembly and one of the two sliders on the bottom of the head coil section.

[41] **Figure 18** is a bottom isometric view of the head coil section showing the roller assembly and a slider on each side of the bottom of the head coil section.

[42] **Figure 19** is an isometric view of the mirror assembly for the head coil section of the neurovascular array housing.

[43] **Figure 20** is an exploded view of the mirror assembly shown in Figure 19.

[44] **Figure 21** illustrates the neurovascular array in a high resolution brain mode.

[45] **Figure 22** illustrates the neurovascular array in an anterior neck mode.

[46] **Figure 23** illustrates the neurovascular array in a C-spine mode.

[47] **Figure 24** illustrates the neurovascular array in a volume neck mode.

[48] **Figure 25** illustrates the neurovascular array in a neurovascular mode.

DETAILED DESCRIPTION OF THE PRESENTLY PREFERRED AND ALTERNATIVE EMBODIMENTS OF THE INVENTION

[49] The presently preferred and alternative embodiments and related aspects of the invention will now be described with reference to the accompanying drawings, in which like elements have been designated where possible by the same reference numerals.

[50] Figure 2 is a schematic diagram for a presently preferred embodiment of a tapered head coil, in the form of a birdcage-type structure, for a neurovascular array (NVA). The head coil 2000 is a "receive-only" resonator, i.e., it does not apply the RF excitation pulses. It may, however, be configured to operate as a transmit/receive (T/R) coil. When configured as a receive-only resonator, the head coil 2000 shall be used with an external transmit coil, such as the RF body coil of the host MR system.

[51] The tapered head coil resonator 2000 comprises two electrically conductive end rings 2101 and 2102, one smaller than the other, interconnected by a plurality of conductive rods A-H. The first and second conductive rings 2101 and 1102 form the inferior and superior ends, respectively, of the head coil 2000. The first conductive ring 2101 has a first diameter, and the second conductive ring 2102 has a second diameter different from the first diameter. The first conductive ring 2101 may, for example, have a diameter of approximately 11.25 inches, and the second conductive ring 2102 a diameter of approximately 5.875 inches. Preferably spaced at irregular distances from each other as shown, for example, in Figures 3 and 10 and described below, the conductive rods A-H extend from the first end ring 2101. More specifically, each of the conductive rods A-H preferably comprises a linear portion and a tapered portion, with the linear portions of the rods being connected to the first end ring 2101. Extending from the linear portions, the tapered portions connect to the second end ring 2102. The tapered portion of rods A-H may be formed from at least one angled linear segmented section. In addition, the length of the conductive rods A-H may, for example, span approximately 9.25 inches.

[52] The exact dimensions for the diameters of end rings 2101 and 2102 and for the lengths of the linear and tapered portions of conductive rods A-H can be selected, of course, to suit the particular use to which the invention will be applied. Specifically, the dimensions for the components of the tapered head coil resonator 2000 should be selected to make the coil particularly useful for imaging any regions of interest encompassed by the coil (e.g., all or any part of the human head). The dimensions cited in U.S. Patent 6,356,081 to *Misic* and U.S. Patent 6,344,745 to *Reisker et al*, for example, will yield improved homogeneity, particularly toward the superior end of the head coil 2000. One or both of the end rings 2101 and 2102, for example, may be circular or elliptical in shape. One or both end rings may also have a diameter larger or smaller than the diameter of the center of the head coil 2000. The spacing of the rods A-H from each other is also a factor, as it affects the capability of head coil 2000 to detect the MR signals emitted from tissue at the center of the imaging volume. With regard to such spacing of the rods, the ability of head coil 2000 to penetrate to the center of the imaging volume is discussed further below.

[53] The head coil 2000 includes both decoupling networks and input resonant circuits. In the presently preferred embodiment in which eight conductive rods A-H are used, the head coil 2000 contains eight input resonant circuits 2111-2118 (one for each mode) and eight decoupling networks 2151-2158 (one for each rod).

[54] Each decoupling network 2151-2158 contains both active and passive decoupling circuits, one acting as a backup for the other. Only decoupling network 2151 is discussed herein in detail, however, because it is representative of the others 2152-2158. As shown in Figure 2, the active decoupling circuit in network 2151 includes PIN diode D_{1A}, diode D_{2A}, variable inductor L_{1A}, and capacitor C_{3A}. During the transmit cycle (i.e., when the RF body coil is transmitting), the host MR system sends a bias signal (e.g., 250 mA) to PIN diode D_{1A}, thus placing both it and diode D_{2A} in a state of forward conduction. This leaves capacitor C_{3A} and variable inductor L_{1A} in parallel, with the equal capacitive and inductive reactances giving rise to a parallel resonant circuit. The resulting high impedance effectively open-circuits that portion of rod A, thus decoupling the entirety of rod A from the host MR system.

During the receive cycle, PIN diode D_{1A} and diode D_{2A} therewith are biased off. The anode of PIN diode D_{1A} of the active decoupling circuit thus sees only the reflected low impedance of its corresponding preamplifier, which can be offered as part of the present invention or be a preexisting part of the host MR system.

[55] The passive decoupling circuit in decoupling network 2151 comprises a pair of parallel-connected diodes D_{3A} and D_{4A} in series with capacitor C_{2A}, both of which in parallel with variable inductor L_{1A} and capacitor C_{3A}. During the transmit cycle, the diodes D_{3A} and D_{4A} respond passively (to the RF signal transmitted by the body coil) by effectively short circuiting themselves. This leaves capacitor C_{3A} and variable inductor L_{1A} in parallel with each other, and again compels them to form a parallel resonant circuit to assure that rod A is indeed open-circuited during the transmit cycle. During the receive cycle of the host MR system, the diode pair D_{3A} and D_{4A} exhibit a high impedance, thereby effectively placing variable inductor L_{1A} in an open circuit. Consequently, in the passive decoupling circuit of network 2151, only the capacitor C_{3A} is seen in rod A during the receive cycles.

[56] Each conductive rod A-H also includes a tuning circuit 2161-2168 in series with its corresponding passive decoupling network 2151-2158. Only tuning network 2161 is discussed herein in detail, however, because it is representative of the others 2162-2168. As shown in Figure 2, the tuning circuit 2161 includes a fixed capacitor C_{4A} and a variable capacitor C_{5A}. As is well known in the art, a head coil can be tuned to optimize its operation by varying the capacitance of its the rods via variable capacitors. The total adjustable range of capacitance in tuning circuit 2161 is preferably 121-136 pF. Therefore, if capacitor C_{4A} is 120 pF then the range of variable capacitor C_{5A} in parallel with C_{4A} would be 1-16 pF, as shown in Figure 2.

[57] The input resonant circuits 2111-2118 are located in the small end ring 2102, with one input resonant circuit being situated between each pair of rods. These define the eight modes of head coil 2000 for purposes of acquiring MR signals from the eight portions of the head using parallel imaging techniques for which the invention is designed. Only input resonant circuit 2111 is discussed herein in detail, though, because it is representative of the others 2112-2118. As shown in Figure 2, input

resonant circuit 2111 includes a port connector $J1_{MI}$, a capacitor C_{4MI}, and a variable inductor L_{4MI}. These components among others enable input resonant circuit 2111 to form a parallel resonant circuit in response to the MR signals it receives during each receive cycle from tissue within its field of view, as is explained further below. The port connector $J1_{MI}$ is used to connect the input resonant circuit to its corresponding preamplifier and, ultimately, to one of the eight channels of the host MR system. Input resonant circuit 2111 also includes other circuitry for, among other purposes, enabling resonance to be achieved for its operating mode as described hereinafter and matching its impedance effectively with that of its corresponding preamplifier.

[58] To render head coil 2000 capable of being used by parallel imaging compatible MR systems, the rods A-H and the first and second end rings 2101 and 2102 have been configured to form a plurality of electrically-adjacent primary resonant substructures about the birdcage-type structure. Essentially forming one coil element, each primary resonant substructure includes two rods and the corresponding sections of the first and second end rings interconnecting them. For the Port I coil element, the primary resonant substructure includes rods H and A and the sections of end rings 2101 and 2102 interconnecting them. For the Port II coil element, the primary resonant substructure includes rods A and B and the corresponding end ring sections interconnecting them. Likewise, the Port III primary resonant substructure includes rods B and C and the sections of end rings interconnecting them, and the Port IV primary resonant substructure includes rods C and D and the corresponding sections of end rings 2101 and 2102. For the Port V coil element, the primary resonant substructure includes rods D and E and the sections of end rings interconnecting them. For the Port VI coil element, the primary resonant substructure includes rods E and F and the corresponding end ring sections, and the Port VII primary resonant substructure includes rods F and G and its corresponding end ring sections. Lastly, for the Port VIII coil element, the primary resonant substructure includes rods G and H and the corresponding sections of end rings 2101 and 2102.

[59] In its presently preferred embodiment, head coil 2000 thus has eight primary resonant substructures, each being sensitive to receiving MR signals from the portion

of the head or other sampled region with its field of view. Used with a parallel imaging compatible MR system, the head coil 2000 is thus capable of providing the MR system simultaneously with the MR signals detected by each of its primary resonant substructures/coil elements. All of the primary resonant substructures are tuned to resonate at the Larmor frequency (i.e., 63.87MHz for 1.5T MR systems).

[60] Furthermore, as shown in Figures 3 and 10-18, the rods A-H of head coil 2000 are preferably spaced at irregular distances from each other. It was found that the ability of the head coil to penetrate to the center of the imaging volume was enhanced by bringing rods A & B closer together, as well as rods C & D, rods E & F and rods G & H. Figure 3, in particular, illustrates a preferred angular distribution of the rods wherein four of the coil elements have their rods spaced 30° apart and the other four have their rods spaced 60° apart. The Port I, III, V, and VII coil elements thus cover twice the area of the Port II, IV, VI and VIII coil elements. In addition to the two sizes, the coil elements are deployed in alternating fashion so that each large one is adjacent to a small one. Despite the irregular spacing of the rods, the symmetry of the head coil is maintained, with respect to the anterior and posterior directions.

[61] The large end ring 2101 also affects resonance for the eight coil elements of head coil 1000. There are some differences in capacitance within the inferior end ring 1101 of head coil 2000. Due to their longer length, the sections of large end ring 2101 corresponding to the larger coil elements have different capacitive values than the sections of large end ring 2101 corresponding to the smaller coil elements. The different capacitive values are used to accommodate the different lengths in the sections of the end ring 2101 between adjacent rods, as best shown in Figures 3 and 10-18. Due the desire to make room for the nose, the section of ring 2101 between rods H & A, for example, is longer than the section opposite it, namely, the section between rods D & E. Similarly, the section of ring 2101 between rods H & A is longer than the adjacent section between A & B. Each of the different capacitive values provides a capacitive reactance that accommodates the different inherent inductances of the the respective sections of the conductive end ring 2101. This compensation scheme was implemented in the large and small end rings because the

length of the rods were chosen to have the same reactive length. Furthermore, as can be seen in Figures 3 and 10-18 in view of Figure 2, the longer sections of ring 2101 (i.e., between rods B & C, D & E, F & G and H and A) have an additional series capacitor, rather than just the fixed and variable capacitors deployed in parallel, for purposes of diminishing the electric field. This arrangement prevents the head coil from being loaded unnecessarily high at the inferior end near the shoulders, and thus avoids a reduction in the overall sensitivity of the head coil 2000. This minimizes the electric field patient coupling to the coil.

[62] The large end ring segment between rods H and A is designed to be resonant generally at the Larmor frequency, and has a small net inductance. Each of the rods H and A is basically series resonant so they have close to zero impedance and zero phase shift end to end. The small end ring segment between rods H and A is designed to be slightly net capacitive, and this slight net capacitance with the slight net inductance of the large end ring segment makes the entire Port I coil element resonant at the Larmor frequency. With regard to the current flow induced within it during the receive cycle, the Port I coil element thus essentially becomes a series-resonant loop that exhibits close to zero net reactance. The current flowing in the Port I coil element is encouraged to remain within it because all conductive paths outside of it (e.g., adjacent segments, and the entirety, of end ring 2101) have been configured to have a higher impedance.

[63] In each coil element, the small net inductance and small net capacitance of the large and small end ring segments, respectively, coupled with zero impedances of two rods corresponding thereto, enables the current induced by the MR signals to flow within the conductive paths of the coil element. In each coil element, very little current in the large end ring segment corresponding thereto will flow into a neighboring large end ring segment for essentially two reasons. First, the collective impedance of the large end ring will discourage it. Each coil element is designed to resonate at the Larmor frequency, but the large end ring 2101 in its entirety is designed to resonate at a frequency significantly higher than that. To the current flowing in each individual large end ring segment, the other segments collectively present a higher

impedance. This higher impedance compels the current flowing within each coil segment to flow from its large end ring segment up its corresponding rod, thus discouraging flow to the neighboring large end ring segment. Only an insignificant amount of current induced within each coil element will find its way to neighboring large end ring segments. Furthermore, given that the large end ring 2101 is tuned significantly above the Larmor frequency, any parasitic currents flowing in the large end ring 2101 will produce a secondary resonance. This secondary resonant substructure will not adversely affect the operation of head coil 2000, however, as it resonates at a frequency sufficiently away from the Larmor frequency. Second, the open circuits presented by the input resonant circuits 2111-2118, discussed below, also discourage it.

[64] The preamplifier to which the port connector $J1_{M1}$ of input resonant circuit 2111 is connected should provide a very low impedance (e.g., less than 2 ohms) so as to cause inductor L_{1M1} and capacitor C_{4M1} to form a parallel resonant circuit. This makes the Port I coil element, particularly its corresponding small end ring segment, appear as an open circuit to the conductive paths of the adjacent coil elements. Virtually all of the current in the Port I loop will therefore flow through the rod and end ring segments of the coil element and the shield and and center conductors of port connector $J1_{M1}$, and thus through the resonant circuit of the preamplifier to which input resonant circuit 2111 is connected. This will produce a voltage across the FET of that preamplifier, with the voltage being relatively large compared to the current that produced it. This voltage will have been produced almost exclusively from the current flowing within the Port I coil element, with virtually no current coming from the other conductive paths of the head coil.

[65] One aspect of this design that is important to the efficient operation of each coil element is the impedance presented at various points within the circuitry. As noted above, each preamplifier should provide a a very low impedance (e.g., > 2 ohms) to its corresponding coil element. (For this reason, it is preferred that the preamplifiers be placed close to the input resonant circuits 2111-2118, otherwise the impedance presented by preamplifiers located further away would be greater due to the longer

cabling needed to interconnect the preamplifiers and the input resonant circuits. Use of remotely located preamplifiers, such as those made available by the manufacturers of MR systems, is not recommended, as such use would make head coil 200 operate less efficiently.) Conversely, each input resonant circuit 2111-2118 should present a significally larger impedance (e.g., on the order of 100 or so ohms) to its corresponding preamplifier. Furthermore, from the perspective of the FET of the preamplifier, it will preferably see an impedance on the order of 2000 ohms when looking at the resonant circuit of the FET and the input resonant circuit beyond. From the perspective of the resonant circuit of the preamplifier, it will preferably see an impedance on the order of 1 MΩ when looking at its corresponding FET in the preamplifier.

[66] The current from any of the other coil elements or conductive paths thereof is thus pretty much blocked by what is called preamplifier decoupling. In each of the other coil elements, the preamplifier provides a low impedance to its corresponding input resonant circuit 2112-2118. This causes the inductor L_{1MX} and capacitor C_{4MX} (where X is 2-8) in those input resonant circuits to resonant at the Larmor frequency during the receive cycles. This makes the small end ring segments of the Port II-VIII coil elements appear like open circuits to the Port I coil element. Because they each form a high impedance parallel resonant circuit with respect to the Port I coil element, the input resonant circuits 2112-2118 of Ports II-VIII block from Port I any current from their respective coil elements and leave the current in the Port I loop to flow only to its corresponding preamplifier.

[67] Each input resonant circuit thus takes advantage of the small input impedance of the preamplifier to which it is connected, and it is this which allows each of them to resonate at the Larmor frequency. This mechanism is important during the receive cycles of the MR system to maintain isolation between each of the Port I-VIII coil elementss. With respect to each coil element, the input resonant circuits of the other seven coil elements will be open circuited.

[68] The operation of head coil 2000, in this preferred embodiment, is now described. During the transmit cycle (i.e., when the RF body coil is transmitting), the host MR

system turns on each of the active and passive decoupling circuits of decoupling networks 2151-2158. Regarding each of the active decoupling circuits, the host MR system sends a bias signal to the PIN diode D_{1X}, thus placing both it and diode D_{2X} in a state of forward conduction. This leaves capacitor C_{3X} and variable inductor L_{1X} in parallel, with the equal capacitive and inductive reactances giving rise to a parallel resonant circuit. In each rod, the resulting high impedance in its active decoupling circuit effectively open-circuits the rod. Regarding the passive decoupling circuits, the diodes D_{3X} and D_{4X} respond passively (to the RF signal transmitted by the body coil) by effectively short circuiting themselves. In each passive decoupling circuit, this leaves capacitor C_{3x} and variable inductor L_{1X} in parallel with each other, and again compels them to form a parallel resonant circuit to assure that the rod to which they correspond is open-circuited. Thus, in each decoupling network 2151-2158, the active and passive circuits assure that the rods A-H will be decoupled from the MR system during the transmit cycle.

[69] During the receive cycle, PIN diode D_{1X} and diode D_{2X} therewith in each of the active decoupling circuits are biased off. For each active decoupling circuit, the anode of PIN diode D_{1X} thus sees only the reflected low impedance of its corresponding preamplifier. For each passive decoupling circuit of networks 2151-2158, the diode pair D_{3X} and D_{4X} exhibit a high impedance, thereby effectively placing variable inductor L_{1X} in an open circuit. Consequently, during the receive cycles, the impedance seen in each of the rods A-H is then only capacitor C_{3X} of the decoupling networks and capacitors of the tuning circuits 2161-2168, respectively.

[70] The head coil 2000 is thus coupled to the MR system during the receive cycle, with each of its operating loops set up as follows. From the perspective of the Port I, Port I sees Ports II-VIII as being open-circuited/off, i.e., as exhibiting a high impedance. This forces the resonant current I out of rod A to flow to the opposite end ring 2101 then up and through rod H through small end ring 2102 and back to Port I. The current then flows into one of the eight channels (e.g., Port I) of the host 8-channel MR system via port connector $J1_{MI}$. Similarly, from the perspective of Port II, Port II sees Ports I and III-VIII as being off. This forces the resonant current I out of rod

A to flow to the opposite end ring 2101 then up and through rod B through small end ring 2102 and back to Port II. This current then flows to another channel (e.g., Port II) in the host MR system via port connector $J1_{M2}$.. From the perspective of Port III, Port III sees Ports I-II and IV-VIII as being off. This forces the resonant current I out of rod B to flow to the opposite end ring 2101 then up and through rod C through small end ring 2102 and back to Port III. The current then flows to the Port III channel of the host MR system via port connector $J1_{M3}$.. The Port IV-VII coil elements operate in like fashion. Lastly, from the perspective of Port VIII, Port VIII sees Ports I-VII as being off. This forces the resonant current I out of rod G to flow to the opposite end ring 2101 then up and through rod H through small end ring 1102 and back to Port VIII. This current then flows to the Port VIII channel of the host 8-channel MR system via port connector $J1_{M8}$.

[71] In a related aspect, the invention also provides the head and base multiplexing circuitry shown in Figures 6A-6B and 7A-7B. Although it can be used without it, this multiplexing circuitry contemplates use of head coil 2000 with anterior neck and posterior C-spine coils of the type generally well known in the art. Figure 4 illustrates an anterior neck coil that has two coil elements, for covering the right and left lateral portions of the neck. Similarly, Figure 5 shows a poster C-spine coil that has two coil elements, for covering the right and left lateral portions of the spine. The head, neck, and C-spine coils herein presented collectively form a neurovascular array (NVA). The quadrature phased array design of the NVA enables complete imaging of the head and the soft tissues of the neck, cervical and upper thoracic spinal regions. The coverage extends from the vertex of the skull to the aortic arch. The multiplexing circuitry serves as the interface between the host 8-channel MR system and the head, neck, and C-spine coils of the NVA.

[72] The multiplexing circuitry enables the NVA to operate in a plurality of operating modes. Figure 21 shows how the NVA would operate in a high resolution brain mode. Specifically, each of the eight coil elements of head coil 2000 would drive one of the channels of the 8-channel host MR system. The anterior neck and C-spine coils would not be used. Figure 22 shows how the NVA would operate in an

anterior neck mode. In this mode, each of the two coil elements of the anterior neck would drive one of the channels of the 8-channel host MR system. The other six channels would not be used. Figure 23 shows how the NVA would operate in a C-spine mode. Like the anterior neck mode, this mode use only two channels of the host MR system, one for each of the two coil elements of the posterior C-spine coil. Figure 24 shows how the NVA would operate in a volume neck mode. In this mode, two channels of the MR system would be driven by the coil elements of the anterior neck coil, and two other channels would be driven by the coil elements of the C-spine coil. The other four channels would not be used.

[73] Figure illustrates how the neurovascular array would operate in a neurovascular mode. In this mode, the head coil 2000, the anterior neck coil and the C-spine coil are all used. Given that the host MR system may only have 8 channels, combiner circuitry within the interface of Figures 6A-6B and 7A-7B permits the outputs of the coil elements to be selectively combined. For example, head coil 2000 may have its eight outputs reduced to 4 pairs of outputs, with each pair representing the combined outputs from two of the coil elements. The coil elements of the anterior neck and the C-spine coil would then be used to drive the other four channels of the MR system.

[74] Other configurations and operational modes, of course, are contemplated by this invention. For example, the NVA of the present invention contemplates at least one spectroscopy mode in which all coil elements of head coil 2000 will be activated. These operating modes and others can be ascertained by reference to the figures.

[75] In the alternative embodiment of head coil 2000 shown in Figure 8, there would be six electrically-adjacent primary resonant substructures for a 6-rod head coil, with each primary resonant substructure being generally deployed 60 degrees apart. For example, the Port I resonant substructure would include rods F and A and the corresponding sections of the end rings interconnecting them, and the Port II resonant substructure would include rods A and B and the sections of the end rings interconnecting them. Similarly, the Port III resonant substructure would include rods B and C and the sections of the end rings interconnecting them, and so on.

[76] In another embodiment shown in Figure 9, there would be twelve primary resonant substructures for a 12-rod head coil, with each substructure being generally deployed 30 degrees apart. For example, the Port I resonant substructure would include rods L and A and the corresponding sections of the end rings interconnecting them, and the Port II resonant substructure would include rods A and B and the sections of the end rings interconnecting them. Similarly, the Port III resonant substructure would include rods B and C and the sections of the end rings interconnecting them, and the Port IV resonant substructure would include rods C and D and the sections of the end rings interconnecting them. Likewise, the Port XI resonant substructure would include rods J and K and the sections of the end rings interconnecting them, and the Port XII primary resonant substructure would include rods K and L and the sections of the end rings interconnecting them. Given the proximity of the rods, the rods are preferably spaced at regular distances from each other.

[77] The design of head coil 2000 is superior compared to approaches adopted by other manufacturers. MRI Devices Corporation of Waukesha, Wisconsin, for example, has also made a head coil to be used with MR systems capable of parallel imaging. The MRI Devices' head coil features overlapped loops, which provide less coverage of the sensitive region and exhibits a 2 to 1 drop in sensitivity between points near the conductive rods and those at the center of the region of interest. Head coil 2000 of the invention, however, provides greater coverage of the sensitive region. It also exhibits a significantly lower drop in sensitivity between points near the conductive rods and those at the center of the volume. By bringing selected rods closer together as disclosed herein, parasitic effects have been further minimized. This yields an improvement in the signal-to-noise ratio while allowing the head coil to obtain consistently a significantly low drop in sensitivity between points near the rods and those at the center of the imaging volume. The preferred embodiment is thus an advance over prior art head coils, as the ability of the head coil 2000 to penetrate to the center of the imaging volume has been enhanced.

[78] It should also be understood that a preamplifier may optionally be added to each of the input resonant circuits 2111-2118. In this variation, of course, preamplifiers in

the host 8-channel MR system would not be needed. In addition, the invention herein disclosed may be used with the existing 9.X software used with the GEMS Signa® 8-channel 1.5 Tesla MR system.

[79] The tapered head coil resonator 2000 can take form in various configurations of components and component placement. Figure 2, which illustrates a band pass configuration, is for the purpose of illustrating a preferred embodiment and is not to be construed as limiting the invention. In particular, the components alternatively may be selected and placed, in a manner known to those skilled in the art, to create a low pass or high pass configuration of the tapered head coil 2000.

[80] The head coil 2000 may also be configured to be a transmit/receive (T/R) coil. This would require removal from the rods A-H of the active and passive decoupling circuits of decoupling networks 2151-2158. In addition, the transmit power (RF energy) from the transmitter port of the host MR system would have to be properly split and routed to each of the Ports I-VIII of the primary resonant substructures with the appropriate phase during the transmit cycle.

[81] It should be apparent that the head coil of the present invention could also be shaped in the form of cylinder, one having straight or non-tapered rods. In another alternative configuration, the head coil could still have the end rings 2101 and 2102 of different diameters but with straight, rather than tapered, rods interconnecting them. The tapered head coil detailed above is preferred, however, because it provides improved field homogeneity on the XZ and YZ image planes.

[82] As will become apparent from the disclosure below, head coil 2000 may be used with other local coil components to form a neurovascular array. For example, the head coil 2000 can be used with an anterior neck coil and a cervical spine coil in a plug type of arrangement known in the art. In one such configuration, the head coil 2000 of the invention would occupy four channels of the host 8-channel MR system, the anterior neck coil shown in Figure 4 would occupy two channels, and the C-spine coil of Figure 5 the last two channels. Other configurations and operational modes, of course, are contemplated by this invention.

[83] Figures 10-20 illustrate one type of neurovascular array (NVA) housing into which the head coil 2000 of the preferred embodiment can be incorporated. In addition to the head coil section, the NVA housing comprises an anterior neck coil section, a posterior cervical spine (or "C-spine") coil section, and a base section. More particularly, as shown in the exploded views of Figures 11 and 13, the NVA housing basically includes a subhousing for each of its coil sections. The head coil section, for example, has an inner and outer subhousings between which is secured the circuitry of head coil 2000 according to the schematic of Figure 2. Figures 10-12 and 19-20 also show the mirror assembly that connects to the head coil section.

[84] Figures 11 and 12 also show the subhousing(s) to which the circuitry of the posterior C-spine coil can be secured according to the schematic of Figure 5. The base section upon which the C-spine section is fixed is also illustrated in Figures 10-15. The head coil section is slideably attached to the base section by means of a slider channel assembly and a roller assembly. The slider channel and roller assemblies are best shown in Figures 16-18. Figures 11 and 12 further illustrate the outer and inner subhousings respectively, between which the circuitry of the anterior neck coil can be secured according to the schematic of Figure 4. The paddle arms and pivot damper assemblies for the anterior neck section are also shown. Besides allowing the anterior neck section to be pivoted between the fully engaged position shown in Figure 13 and the fully upright position shown in Figure 14, the pivot assemblies each feature a dampening means by which the paddle arms, and the anterior neck section therewith, may be positioned at any point along their range of movement from and in between the fully upright and engaged positions. The pivot assemblies both mount to the base section but on opposite sides of the outer subhousing for the head coil 2000, as best shown in Figures 11 and 12.

[85] The base and cover therefor are also illustrated in Figures 11 and 12. The base serves as the main mounting structure for the other sections of the NVA housing. For example, the bottom of the C-spine section mounts directly to the base. The bottom or posterior portion of the head coil section is also supported by the base. The base cover provides further support for the C-spine section and the head coil section.

[86] The NVA housing of the present invention has been designed to facilitate use by the technologist/operator and to improve the comfort of the patient. The NVA housing is lightweight and has an ergonomic design. During those operation modes in which head coil 2000 is not used, the head coil section of the NVA can be slid away from the patient while the anterior neck and/or posterior C-spine coils are used. Although mechanically present during such procedures, the head coil section is electrically disabled and invisible to the MR system. This unique open architecture, along with the spacing of the rods, tends to minimize claustrophobic reactions. This is quite advantageous for the anterior neck, C-spine, and volume neck modes in which only soft tissues of the neck, cervical and upper thoracic spinal regions are imaged.

[87] The presently preferred and alternative embodiments for carrying out the invention have been set forth in detail according to the Patent Act. Persons of ordinary skill in the art to which this invention pertains may nevertheless recognize alternative ways of practicing the invention without departing from the spirit of the following claims. Consequently, all changes and variations which fall within the literal meaning, and range of equivalency, of the claims are to be embraced within their scope. Persons of such skill will also recognize that the scope of the invention is indicated by the following claims rather than by any particular example or embodiment discussed or illustrated in the foregoing description.

[88] Accordingly, to promote the progress of science and useful arts, we secure for ourselves by Letters Patent exclusive rights to all subject matter embraced by the following claims for the time prescribed by the Patent Act.

CLAIMS

What is claimed is:

1. A birdcage coil for use with a magnetic resonance (MR) system capable of acquiring images of a region of interest using parallel imaging techniques; the birdcage coil comprising:

(a) a first ring at an inferior end of the birdcage coil, said first ring being electrically conductive and having a first diameter through which the region of interest is provided access to the birdcage coil;

(b) a second ring at a superior end of the birdcage coil, said second ring being electrically conductive and having a second diameter smaller than said first diameter of said first ring; and

(c) a plurality of rods electrically interconnecting said first and said second rings to form the birdcage coil therewith, each of said rods having a linear portion and a tapered portion with said linear portion being connected to said first ring and said tapered portion being connected to said second ring, said tapered portions of said rods collectively providing the birdcage coil with a substantially homogeneous pattern of magnetic flux density in at least one of three orthogonal imaging planes of the birdcage coil while at least one of maintaining and improving a signal-to-noise ratio of the birdcage coil;

wherein said rods and said first and said second rings are configured to form about the birdcage coil a plurality of electrically-adjacent primary resonant substructures, with each of said primary resonant substructures including two of said rods and a corresponding section of each of said first and said second rings interconnecting them, which are capable of being used by said MR system for simultaneous reception of magnetic resonance signals from the region of interest.

2. The birdcage coil of claim 1 wherein the birdcage coil is a receive-only coil.

3. The birdcage coil of claim 1 wherein each of said rods and said first and said second rings contain therein a plurality of reactive electrical components.

4. The birdcage coil of claim 1 wherein each of said primary resonant substructures includes a port connector in said corresponding section of said second ring thereof for connection to a channel of the MR system.

5. The birdcage coil of claim 1 wherein the birdcage coil is a transmit/receive coil.

6. The birdcage coil of claim 1 wherein the birdcage coil is configured as one of a low pass coil, a high pass coil and a band pass coil.

7. The birdcage coil of claim 1 wherein said first and said second rings are circular.

8. The birdcage coil of claim 1 wherein at least one of said first and said second rings are elliptical.

9. The birdcage coil of claim 1 further comprising at least one additional coil at least partially overlapping the birdcage coil at the inferior end thereof to extend the field of view.

10. The birdcage coil of claim 1 wherein said first ring and said second ring are each larger in diameter than a center of the birdcage coil.

11. The birdcage coil of claim 1 wherein said rods and said first and said second rings are supported and housed within a housing therefor.

12. A birdcage coil for use with a magnetic resonance (MR) system capable of acquiring images of a region of interest using parallel imaging techniques; the birdcage coil comprising:

(a) a first ring at one end of the birdcage coil, said first ring being electrically conductive and having a first diameter;

(b) a second ring at an other end of the birdcage coil, said second ring being electrically conductive and having a second diameter different from said first diameter of said first ring; and

(c) a plurality of rods electrically interconnecting said first and said second rings to form the birdcage coil therewith;

wherein said rods and said first and said second rings are configured to form about the birdcage coil a plurality of electrically-adjacent primary resonant substructures, with each of said primary resonant substructures including two of said rods and a corresponding section of each of said first and said second rings interconnecting them, which are capable of being used by said MR system for simultaneous reception of magnetic resonance signals from the region of interest.

13. The birdcage coil of claim 12 wherein each of said rods has a linear portion and a tapered portion with said linear portion being connected to said first ring and said tapered portion being connected to said second ring, said tapered portions of said rods collectively providing said coil with a substantially homogeneous pattern of magnetic flux density in at least one of three orthogonal imaging planes of said coil.

14. The birdcage coil of claim 13 wherein said tapered portion of each of said rods comprises at least one angled linear segmented section.

15. The birdcage coil of claim 12 wherein said second diameter of said second ring is smaller than said first diameter of said first ring.

16. The birdcage coil of claim 12 wherein said first and said second rings are circular.

17. The birdcage coil of claim 12 wherein at least one of said first and said second rings are elliptical.

18. The birdcage coil of claim 12 wherein selected ones of said rods are spaced at irregular distances from adjacent ones of said rods.

19. A birdcage coil for use with a magnetic resonance (MR) system capable of acquiring images of a region of interest using parallel imaging techniques; the birdcage coil comprising:

(a) a first ring at one end of the birdcage coil, said first ring being electrically conductive and having a first diameter;

(b) a second ring at an other end of the birdcage coil, said second ring being electrically conductive and having a second diameter; and

(c) a plurality of rods electrically interconnecting said first and said second rings to form the birdcage coil therewith;

wherein said rods and said first and said second rings are configured to form about the birdcage coil a plurality of electrically-adjacent primary resonant substructures, with each of said

primary resonant substructures including two of said rods and a corresponding section of each of said first and said second rings interconnecting them, which are capable of being used by said MR system for simultaneous reception of magnetic resonance signals from the region of interest.

20. A head coil for use with a magnetic resonance (MR) system; the head coil comprising:

(a) a first ring at one end of the head coil, said first ring being electrically conductive and having a first diameter;

(b) a second ring at an other end of the head coil, said second ring being electrically conductive and having a second diameter; and

(c) a plurality of rods electrically interconnecting said first and said second rings to form the head coil therewith;

wherein said rods and said first and said second rings are configured to form a plurality of electrically-adjacent primary resonant substructures in a birdcage-like structure, with each of said primary resonant substructures including two of said rods and a corresponding section of each of said first and said second rings interconnecting them.

ABSTRACT

A head coil for use with a magnetic resonance (MR) system comprises a first ring at one thereof, a second ring at the other end thereof, and a plurality of rods electrically interconnecting the first and second rings. The first ring is electrically conductive and has a first diameter. The second ring is electrically conductive and has a second diameter. The rods and first and second rings are configured to form a plurality of electrically-adjacent primary resonant substructures about a birdcage-type structure. Each primary resonant substructure includes two of the rods and the corresponding sections of the first and second rings interconnecting them.

FIGURE 1A

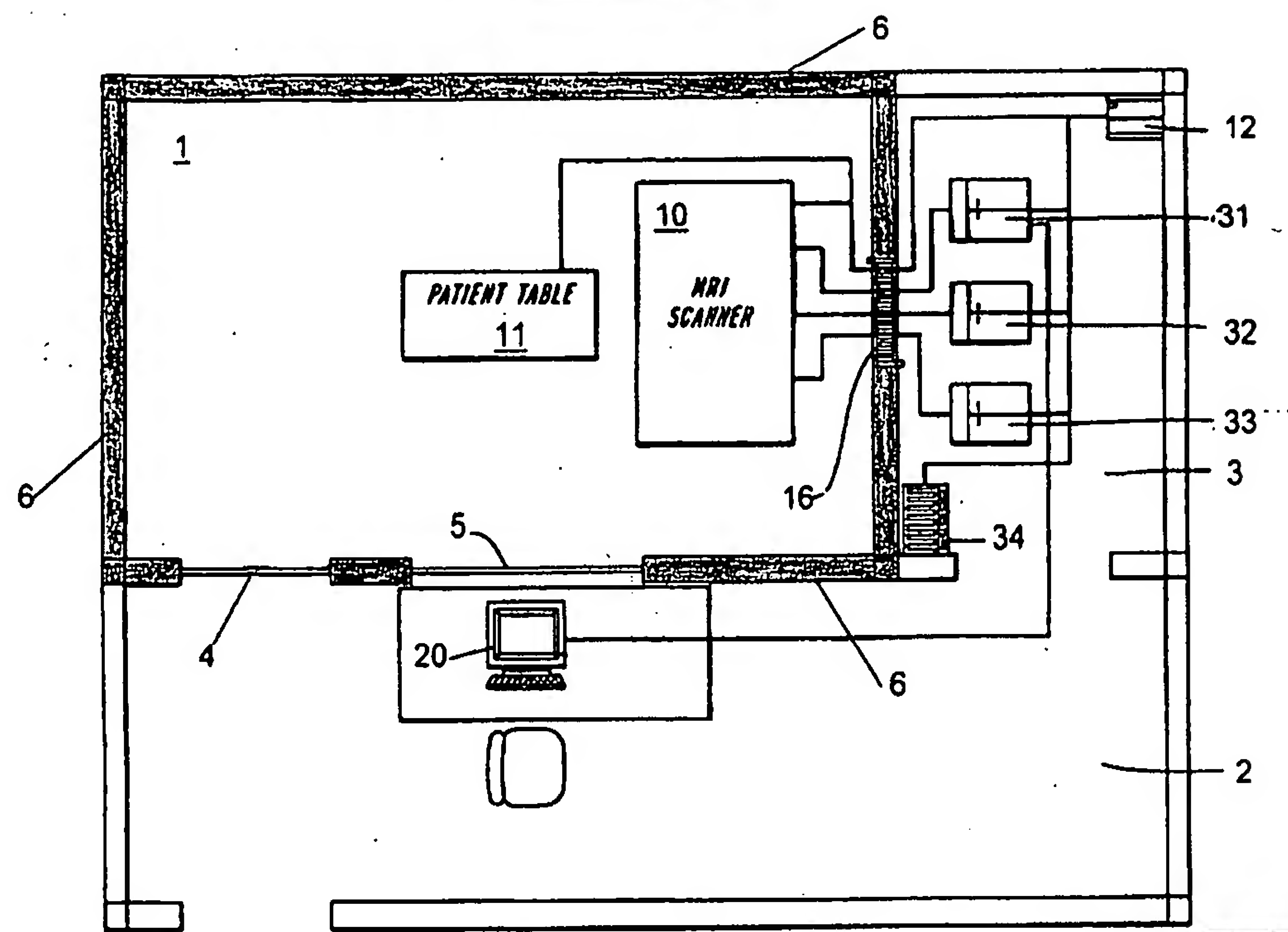


FIGURE 1B

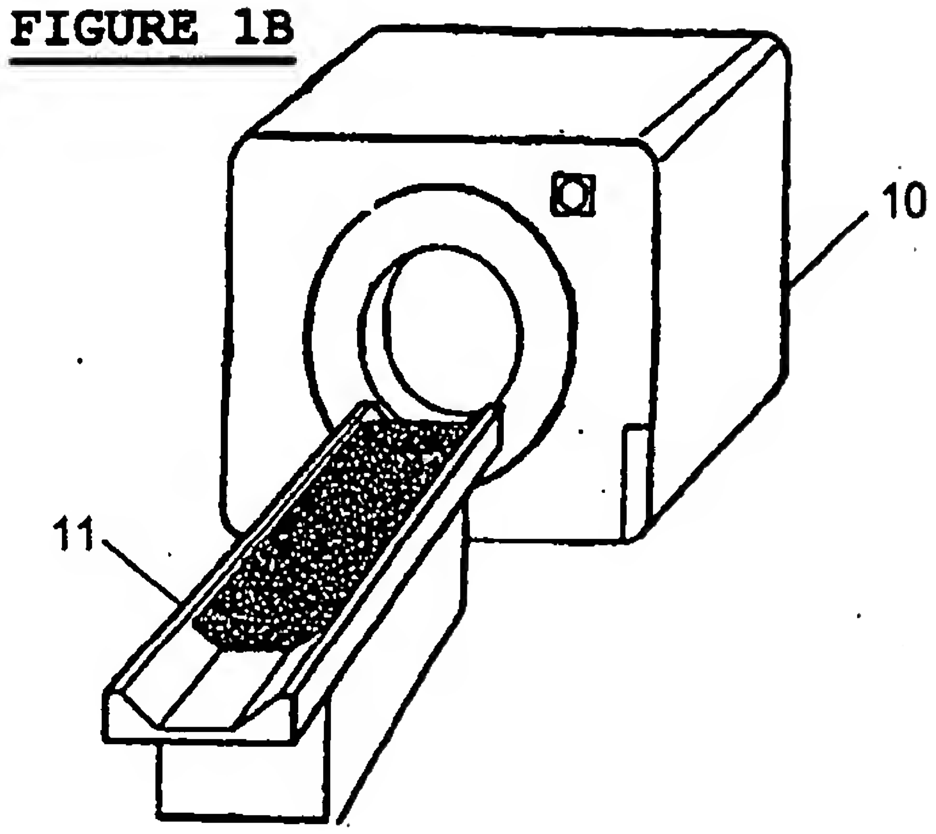


FIGURE 1C

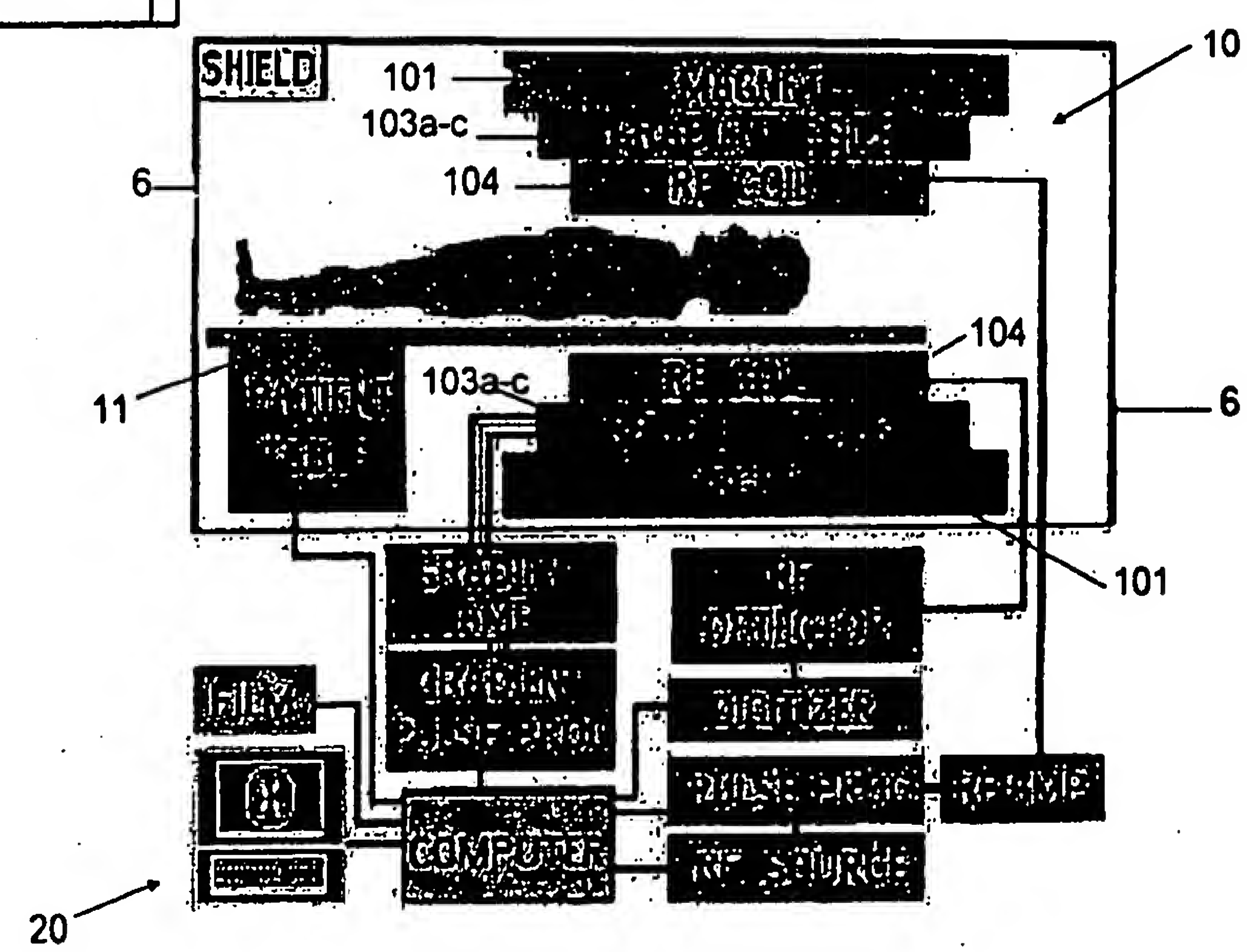
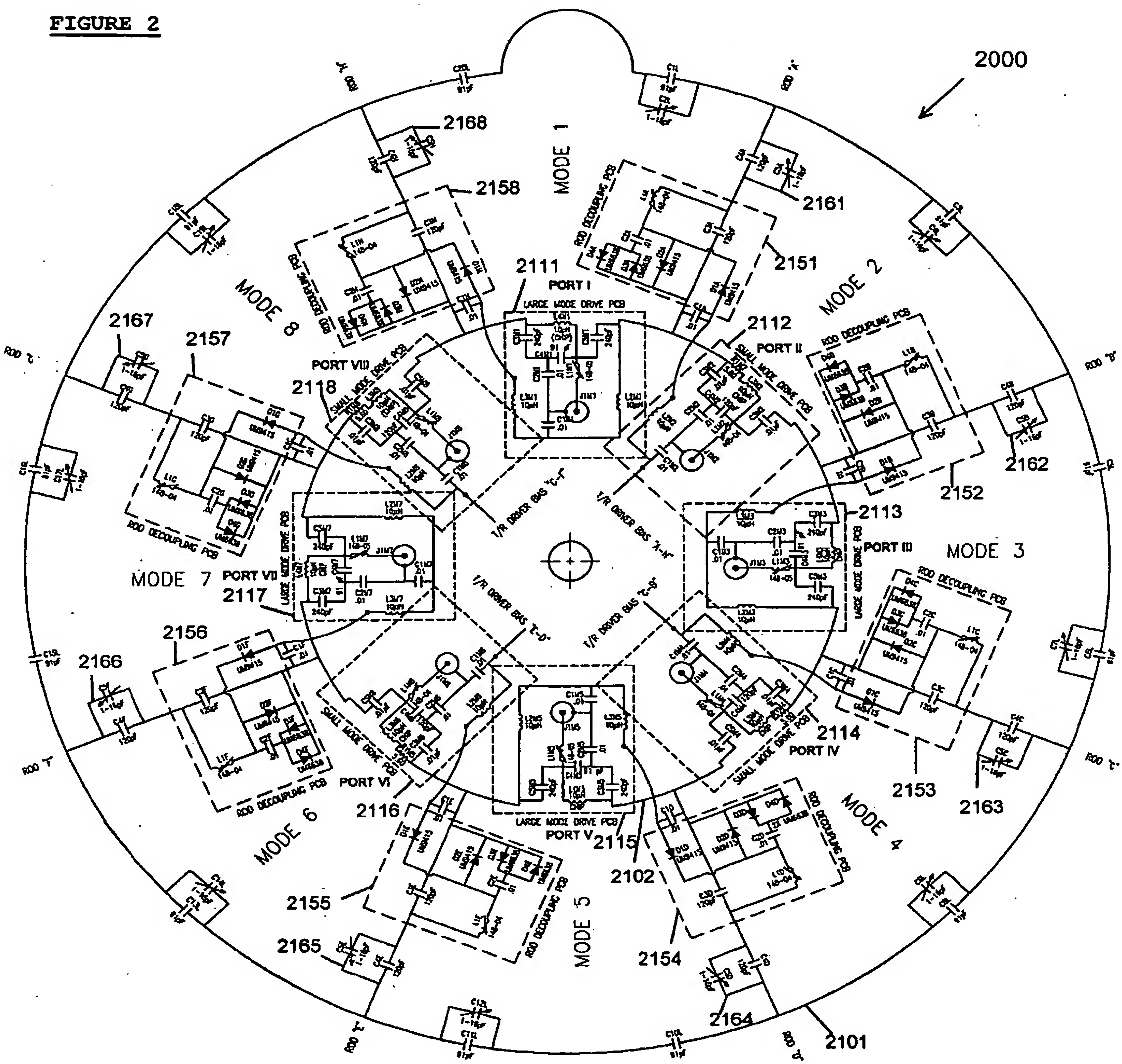


FIGURE 2



2000
MODE 1
MODE 2
MODE 3
MODE 4
MODE 5
MODE 6
MODE 7
MODE 8
PORT I
PORT II
PORT III
PORT IV
PORT V
PORT VI
PORT VII
PORT VIII
LARGE MODE DRIVE PCB
SMALL MODE DRIVE PCB
ROD DECOUPLING PCB
T/R DRIVER BIAS "C-F"
T/R DRIVER BIAS "A-H"
T/R DRIVER BIAS "C-D"
T/R DRIVER BIAS "E-D"
2101
2102
2111
2112
2113
2114
2115
2116
2117
2118
2151
2152
2153
2154
2155
2156
2157
2158
2161
2162
2163
2164
2165
2166
2167
2168

FIGURE 2A

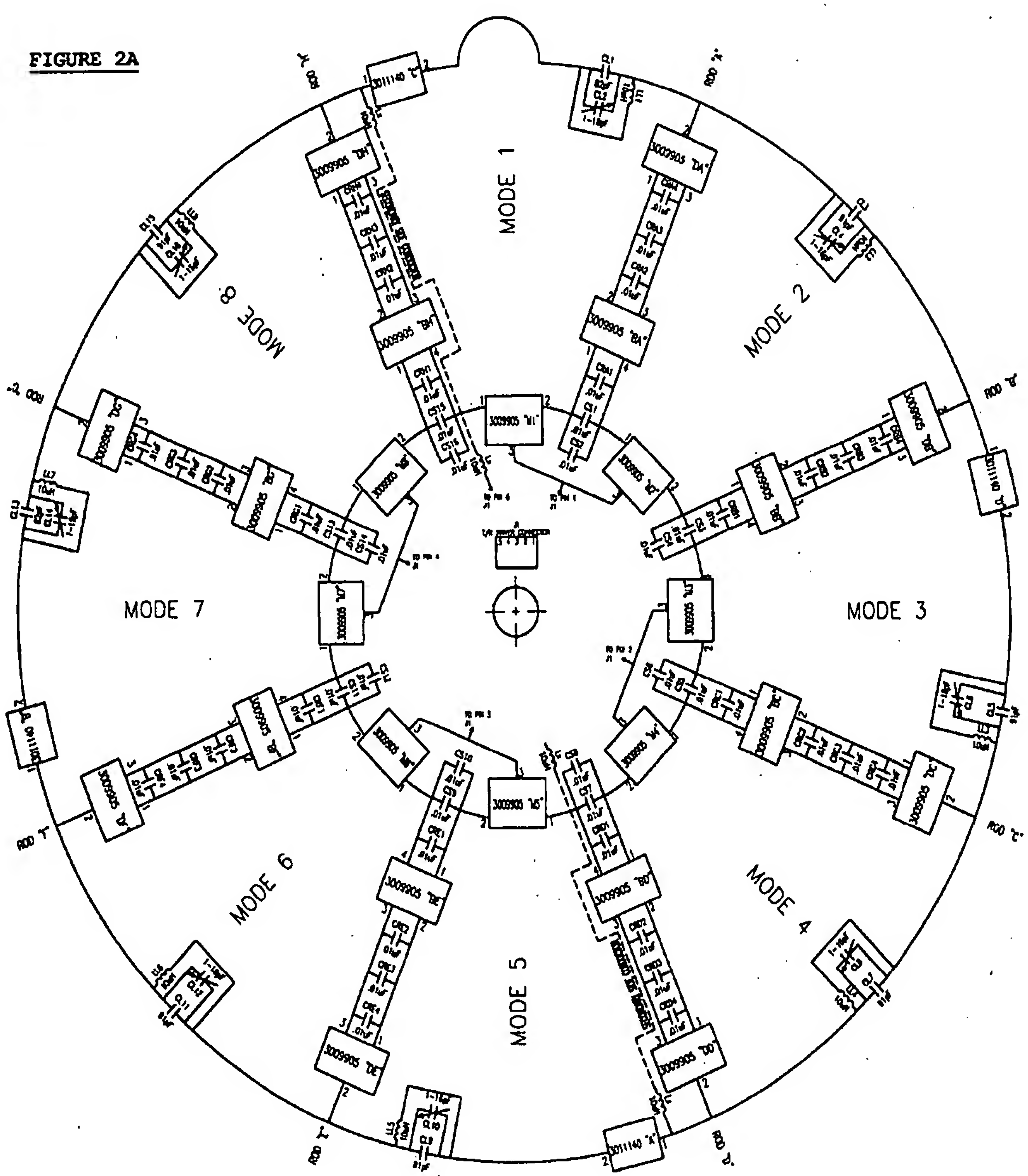
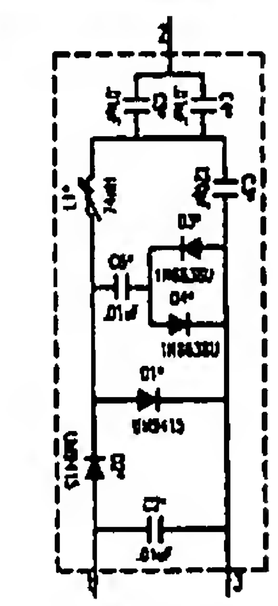
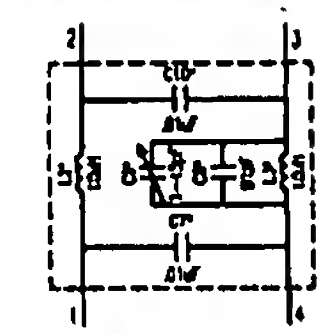


FIGURE 2B



3009905 "D"
"D"="DECOUPLING"
"*"=ROD LOCATION (A THRU H)

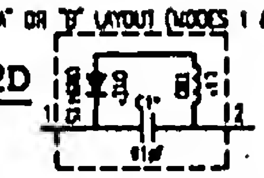
FIGURE 2C



3009905 "B"
"B"="BYPASS"
"*"=ROD LOCATION (A THRU H)

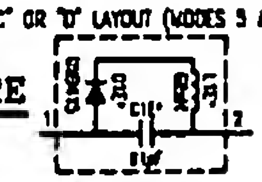
"A" OR "B" LAYOUT (MODES 1 & 3)

FIGURE 2D



"C" OR "D" LAYOUT (MODES 5 & 7)

FIGURE 2E



3011140 "*"
"E"= "END RING PCB" (COMPONENT DESG. ONLY)
"*"= "A" "B" "C" OR "D"

FIGURE 2F

3009905 M*
LARGE MODE DRIVE PCB
"*"=MODES 1, 3, 5 OR 7

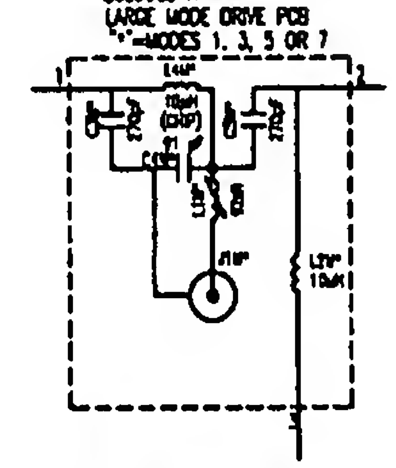
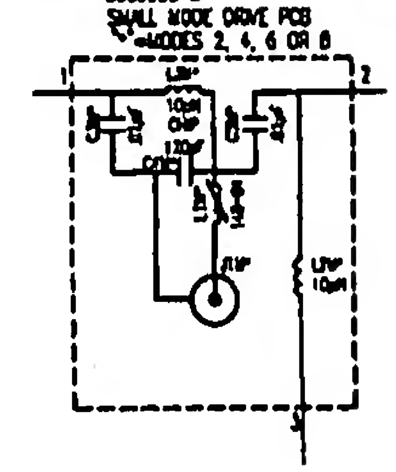


FIGURE 2G

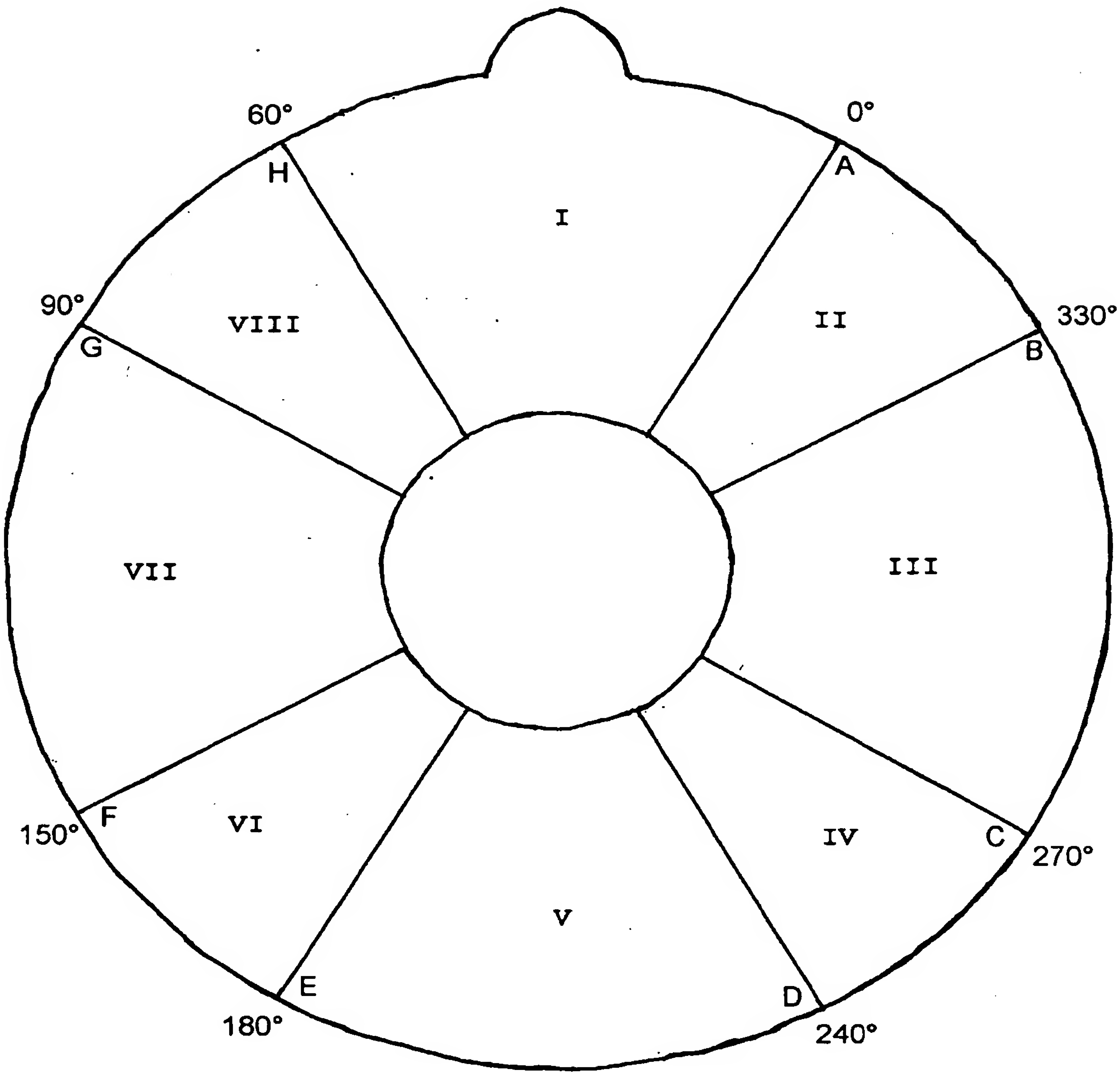
3009905 M*
SMALL MODE DRIVE PCB
"*"=MODES 2, 4, 6 OR 8

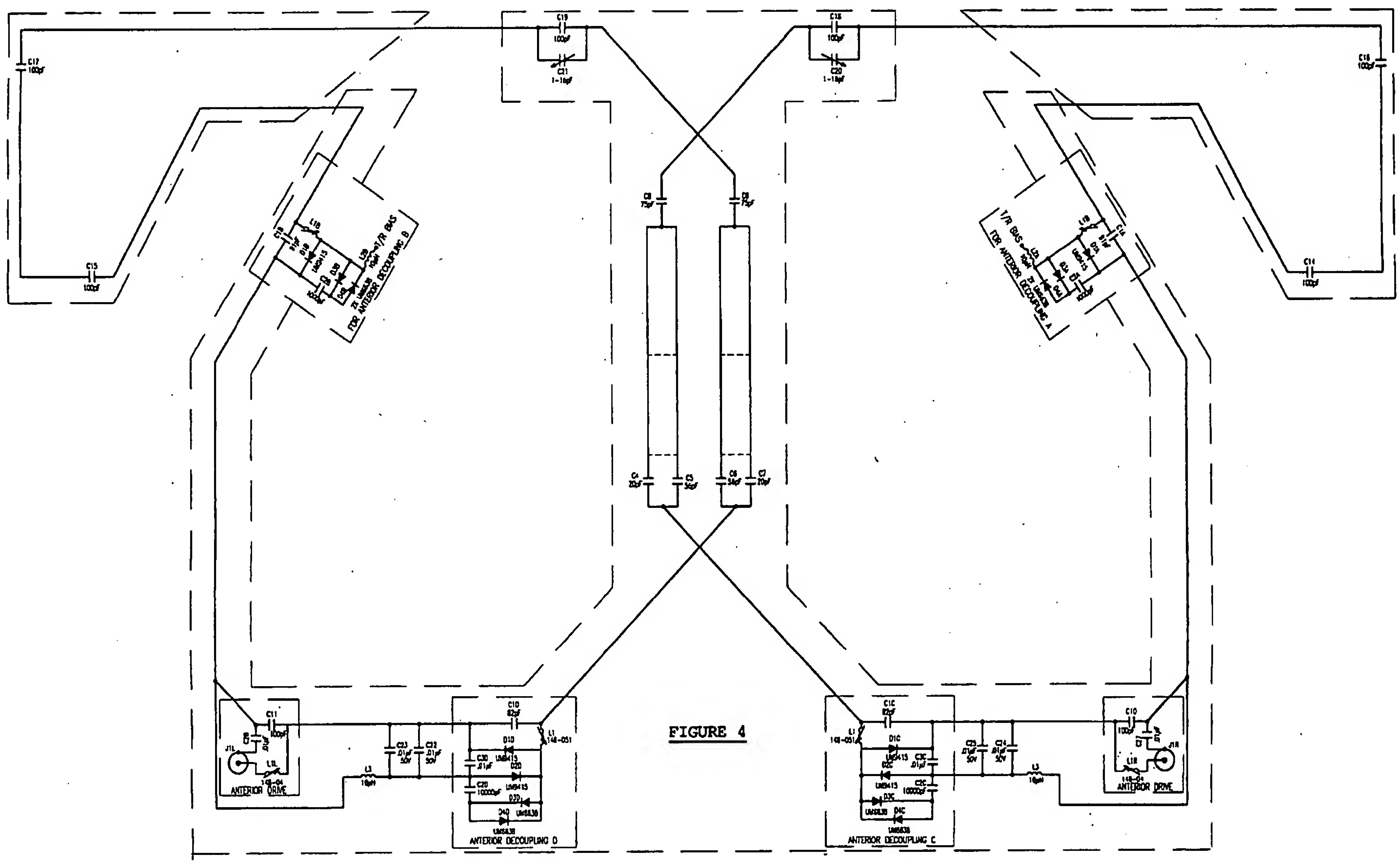


HEAD-COIL SCHEMATIC

(COIL VIEWED FROM SUPERIOR END)

FIGURE 3





C17 100pF
C15 100pF
C19 100pF
C21 1-18pF
C18 100pF
C20 1-18pF
C16 100pF
C14 100pF
C8 75pF
C9 75pF
T/R BIAS
FOR ANTERIOR DECOUPLING B
FOR ANTERIOR DECOUPLING A
ANTERIOR DRIVE
ANTERIOR DECOUPLING D
ANTERIOR DECOUPLING C

FIGURE 4

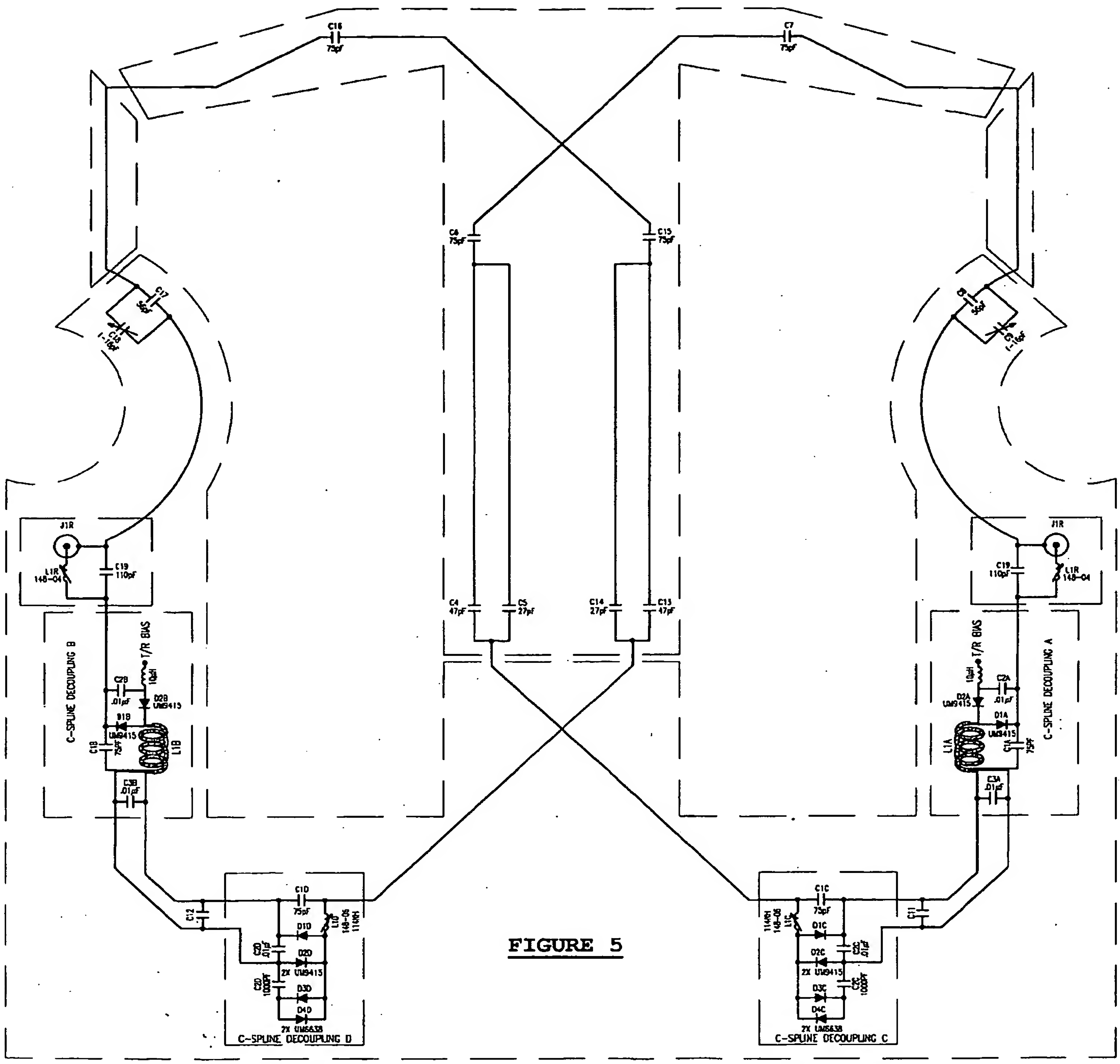


FIGURE 5

C-SPINE SCHEMATIC

FIGURE 6A

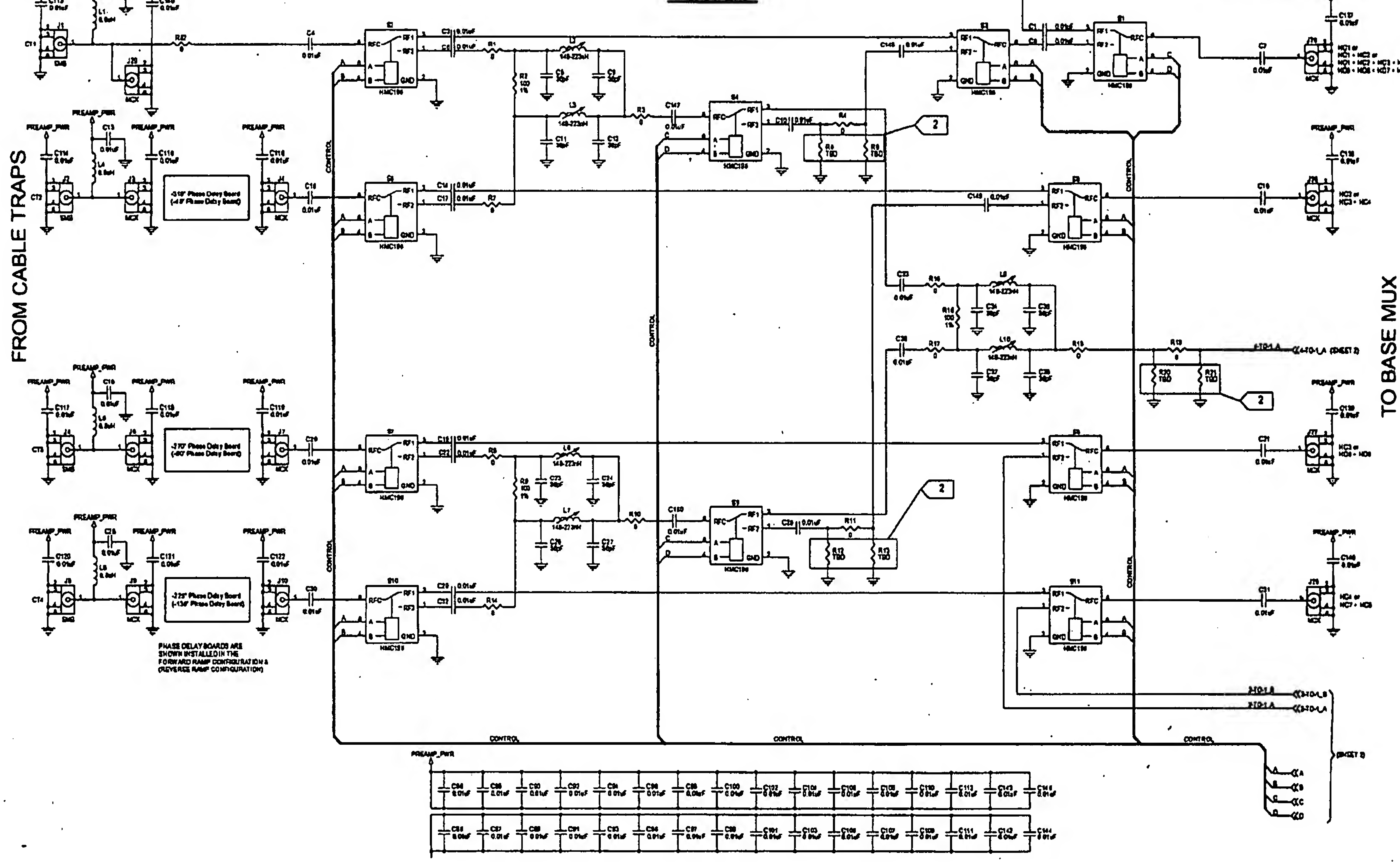

FROM CABLE TRAPS
TO BASE MUX
PHASE DELAY BOARDS ARE SHOWN INSTALLED IN THE FORWARD RAMP CONFIGURATION & (REVERSE RAMP CONFIGURATION)
CONTROL
PREAMP_PWR
HMC199

FIGURE 6B

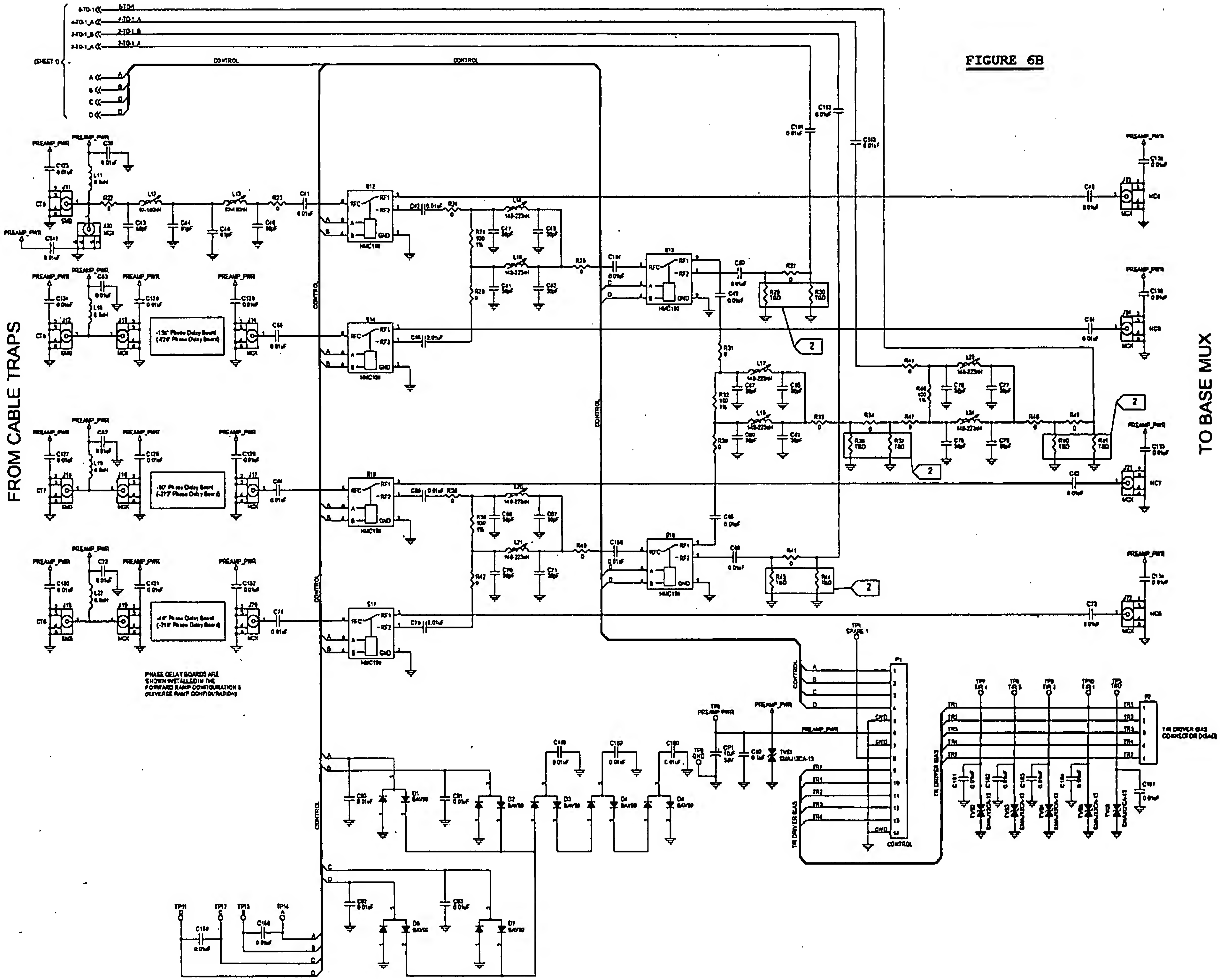

FROM CABLE TRAPS
TO BASE MUX
CONTROL
PREAMP_PWR
PHASE DELAY BOARDS ARE SHOWN INSTALLED IN THE FORWARD RAMP CONFIGURATION & (REVERSE RAMP CONFIGURATION)

FIGURE 7A

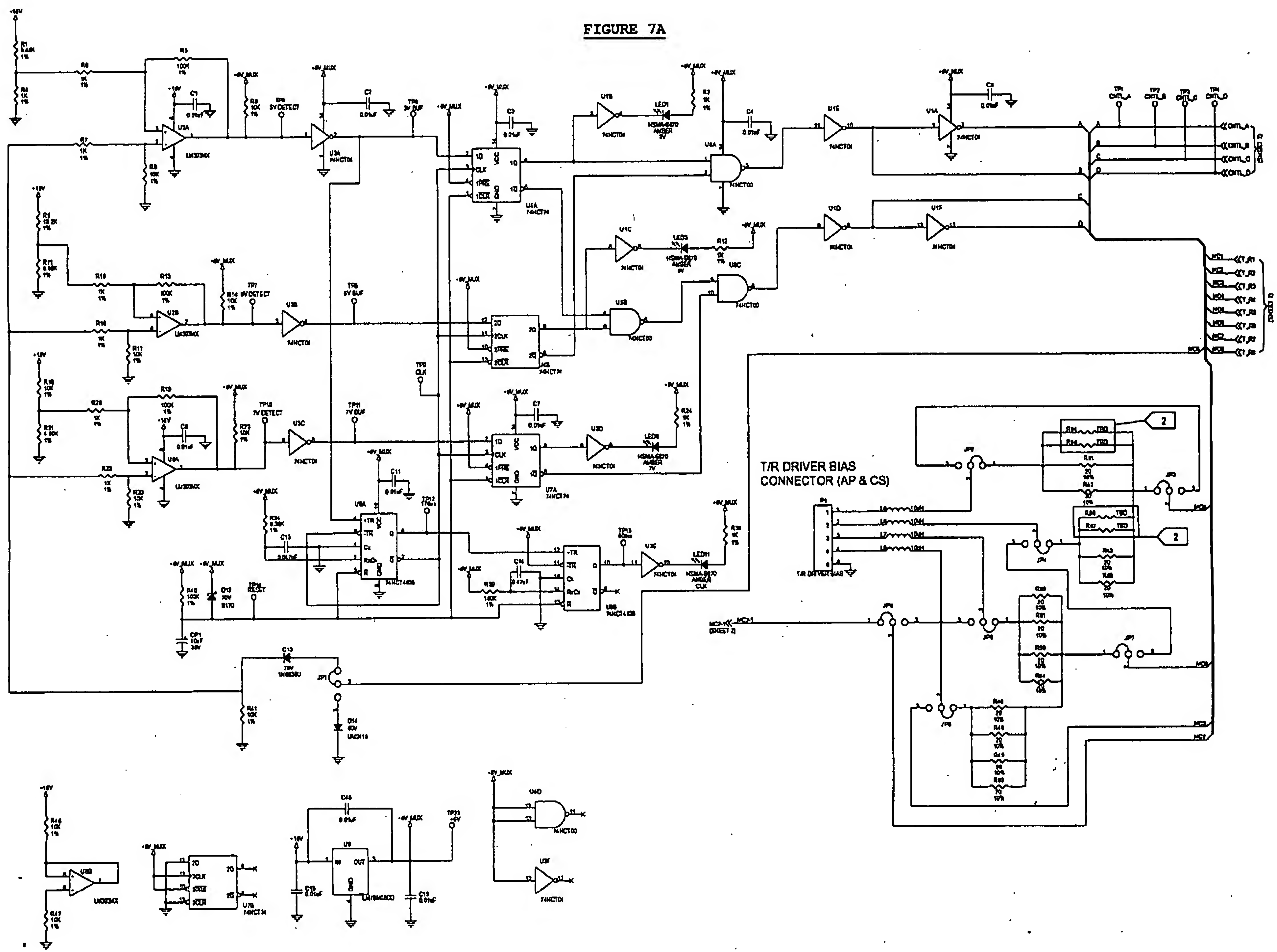
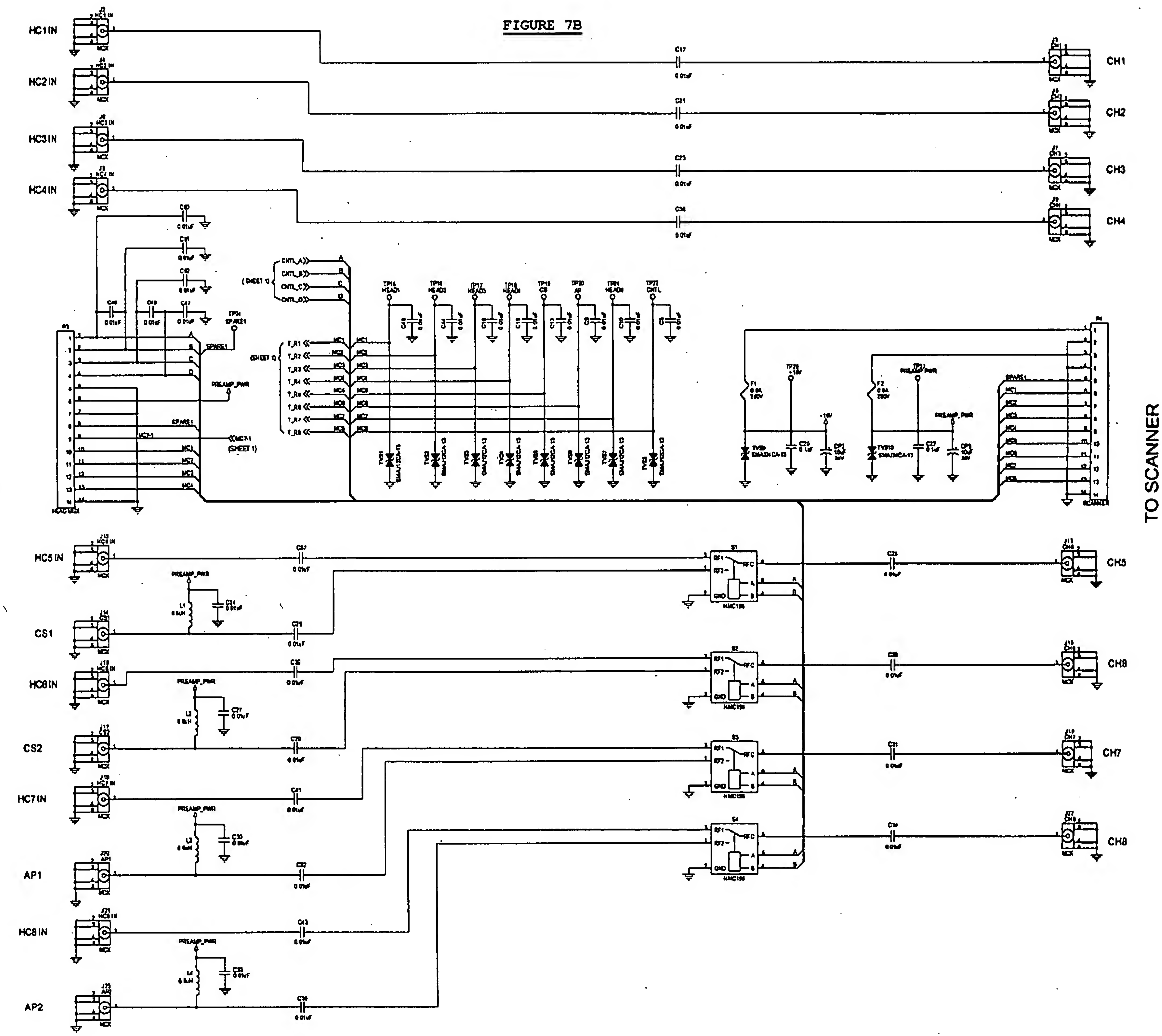


FIGURE 7B



FROM HEAD MUX
TO SCANNER
HC1 IN
HC2 IN
HC3 IN
HC4 IN
HC5 IN
CS1
HC6 IN
CS2
HC7 IN
AP1
HC8 IN
AP2
CH1
CH2
CH3
CH4
CH5
CH6
CH7
CH8

FIGURE 8

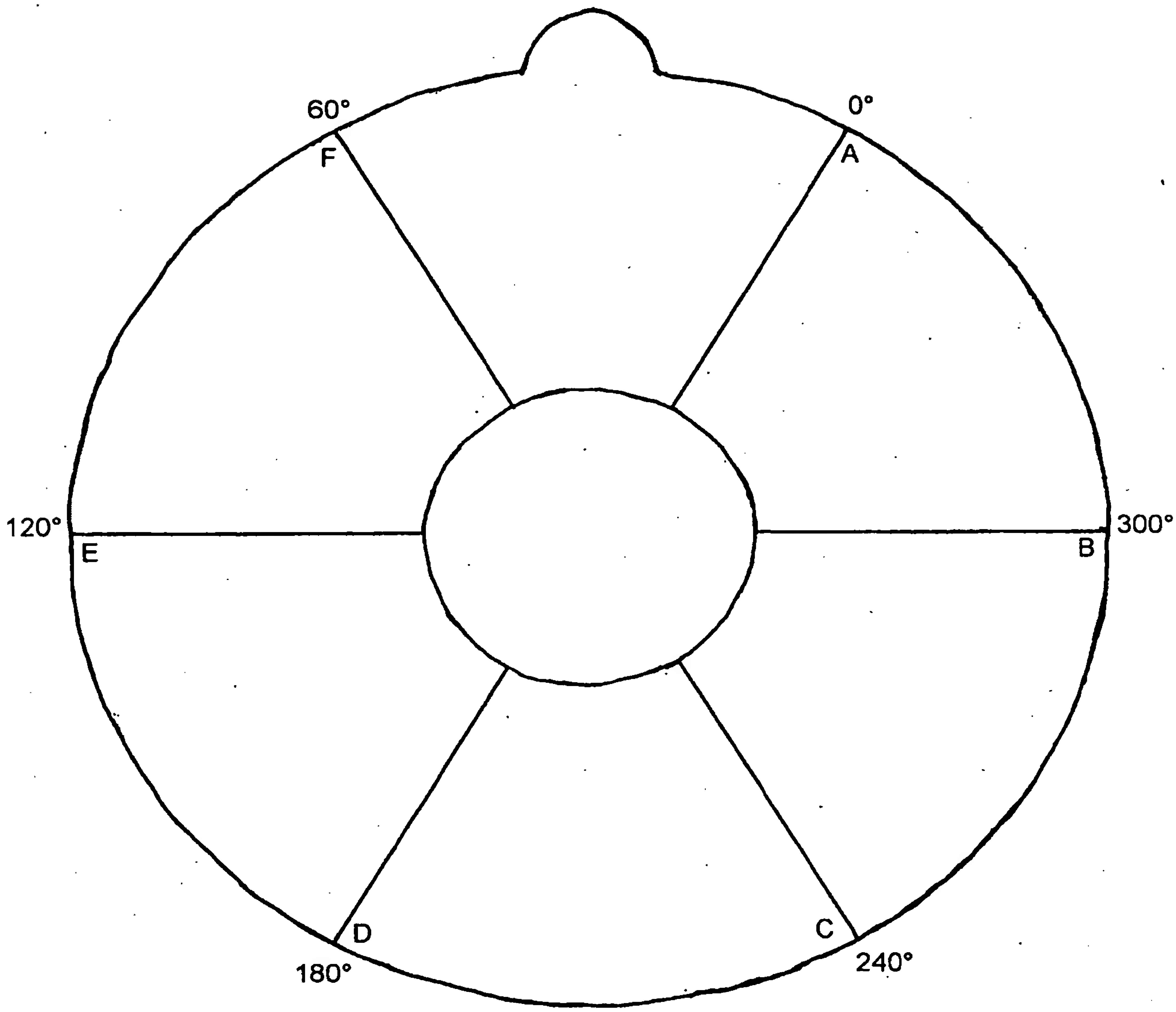


FIGURE 9

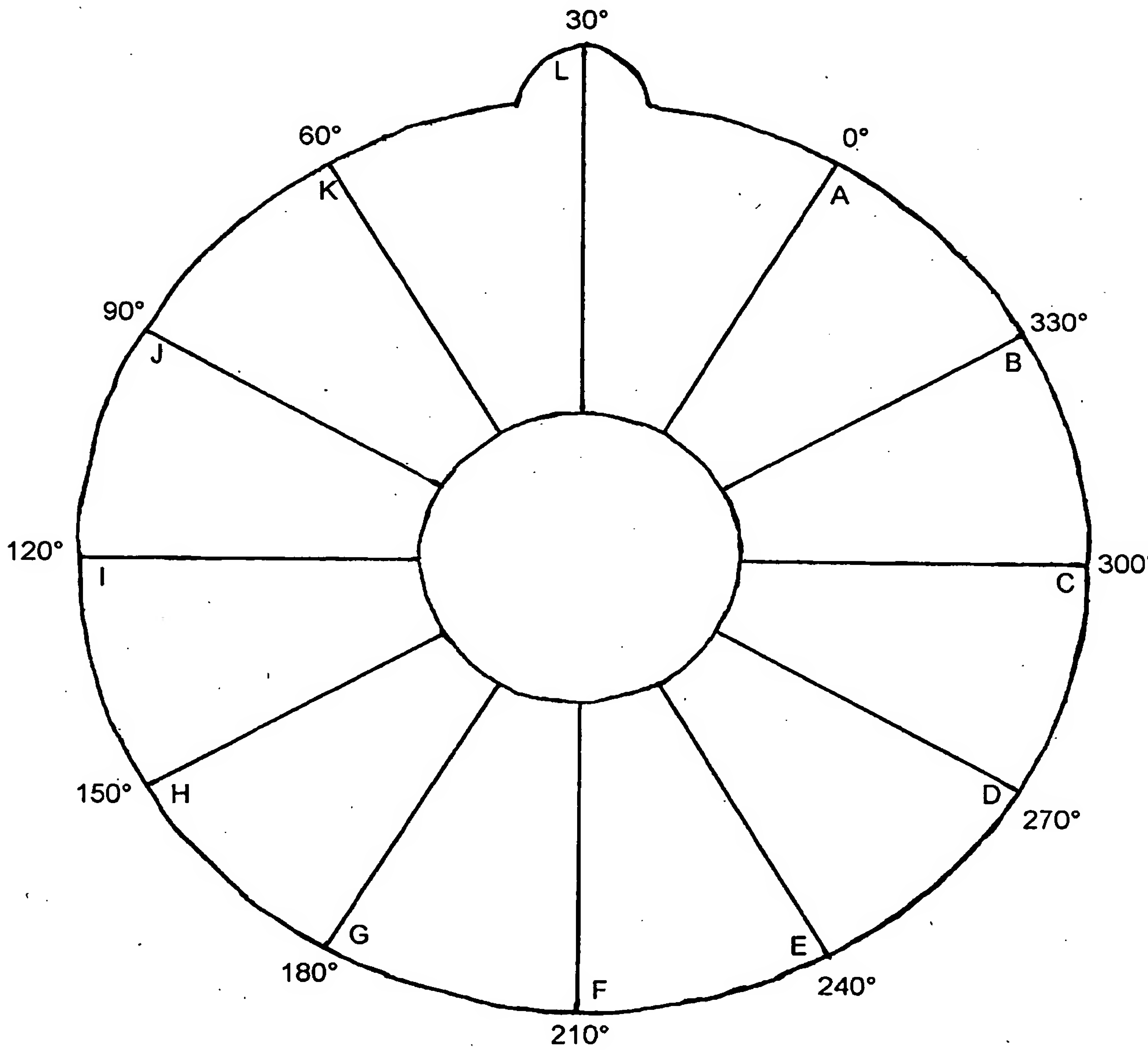


FIGURE 10

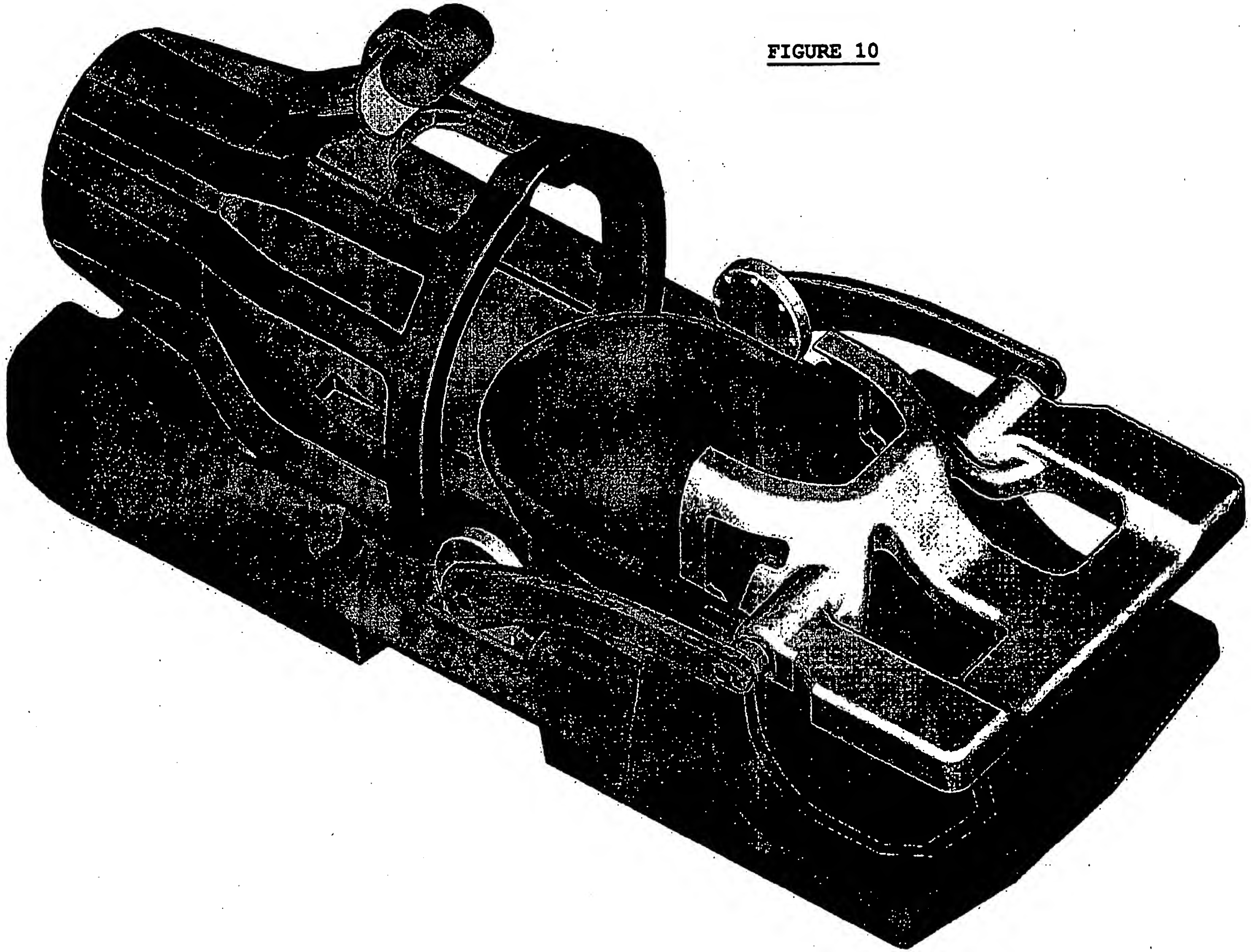


FIGURE 11

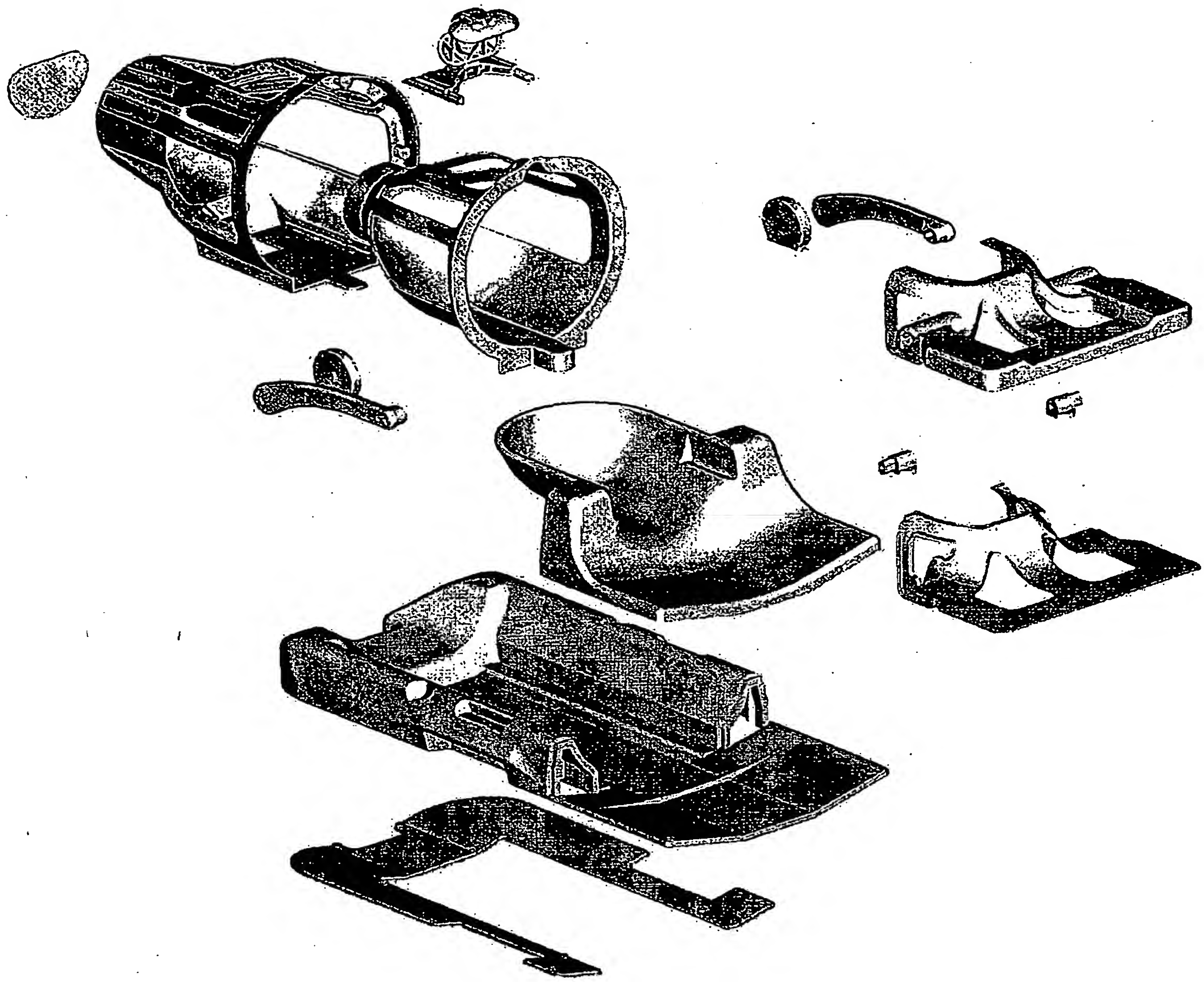


FIGURE 12

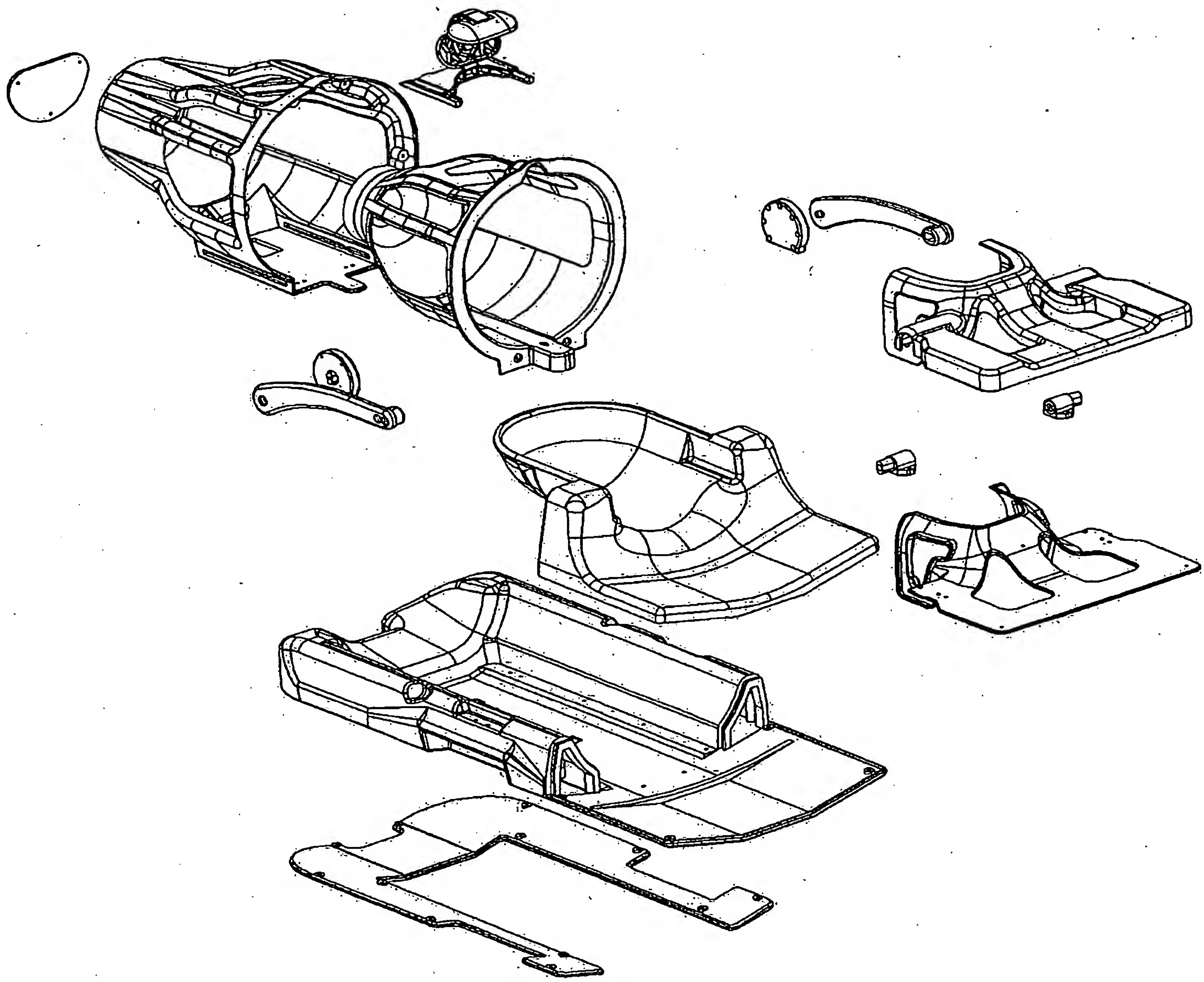


FIGURE 13

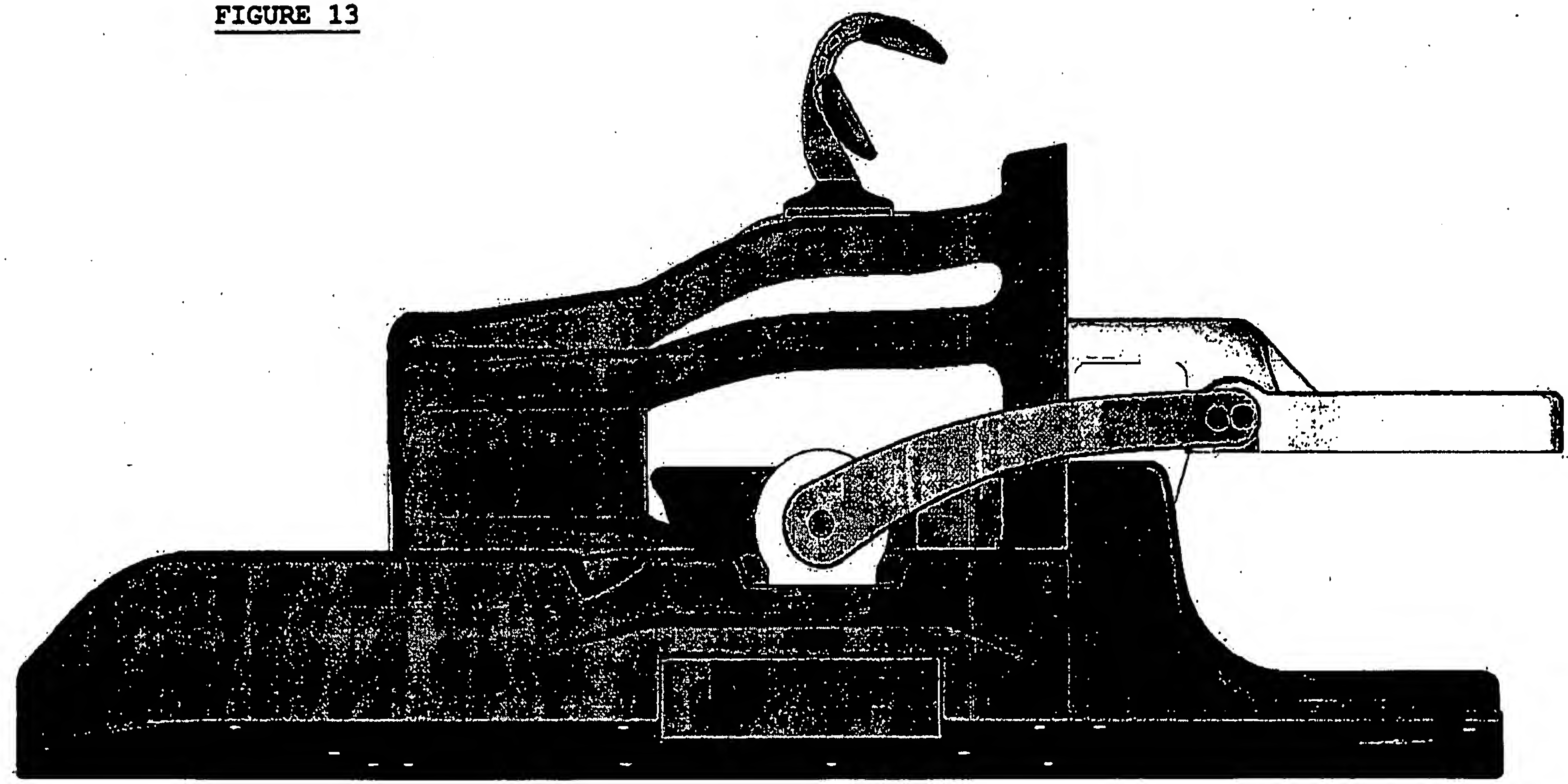


FIGURE 14

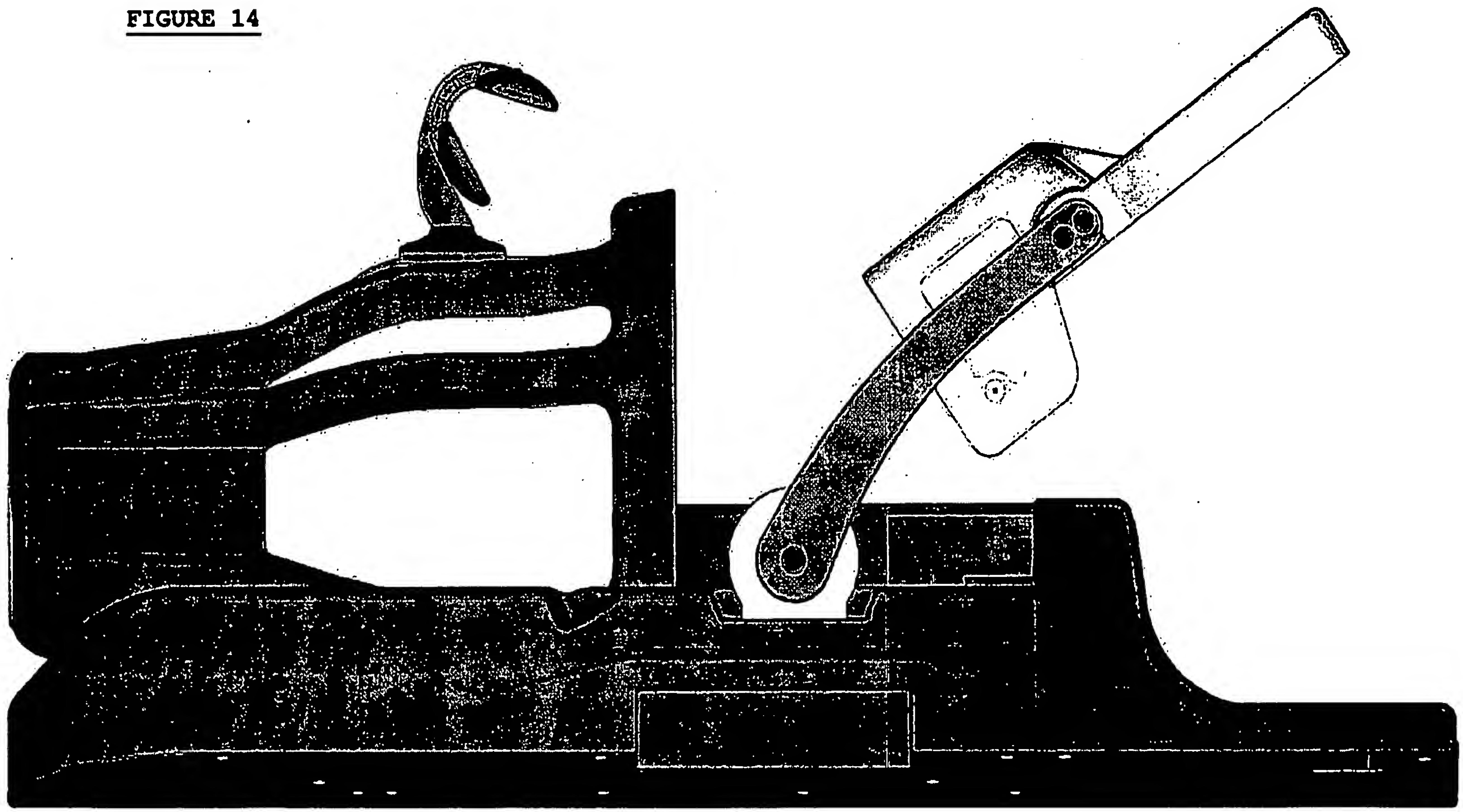


FIGURE 15

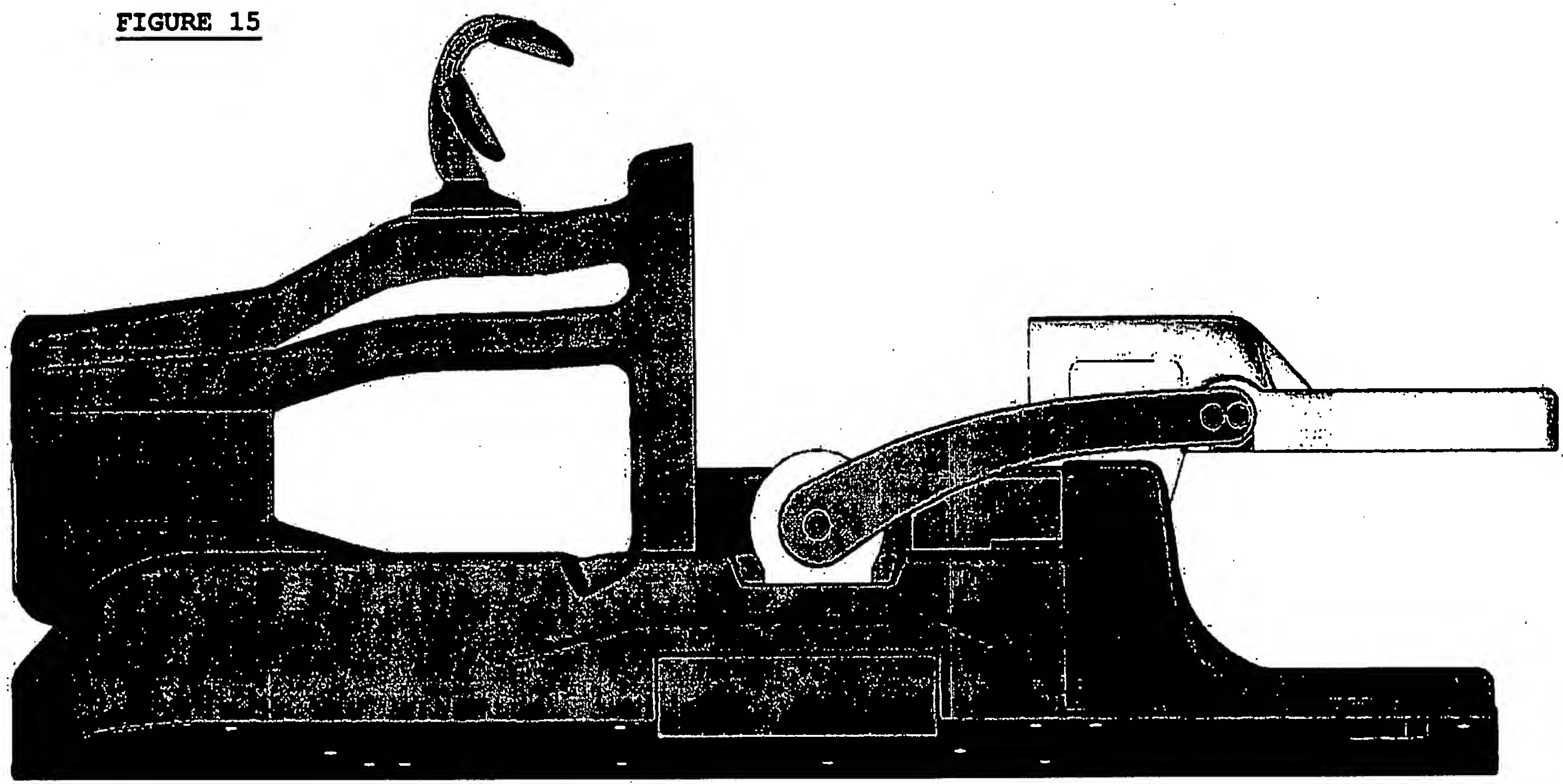


FIGURE 16

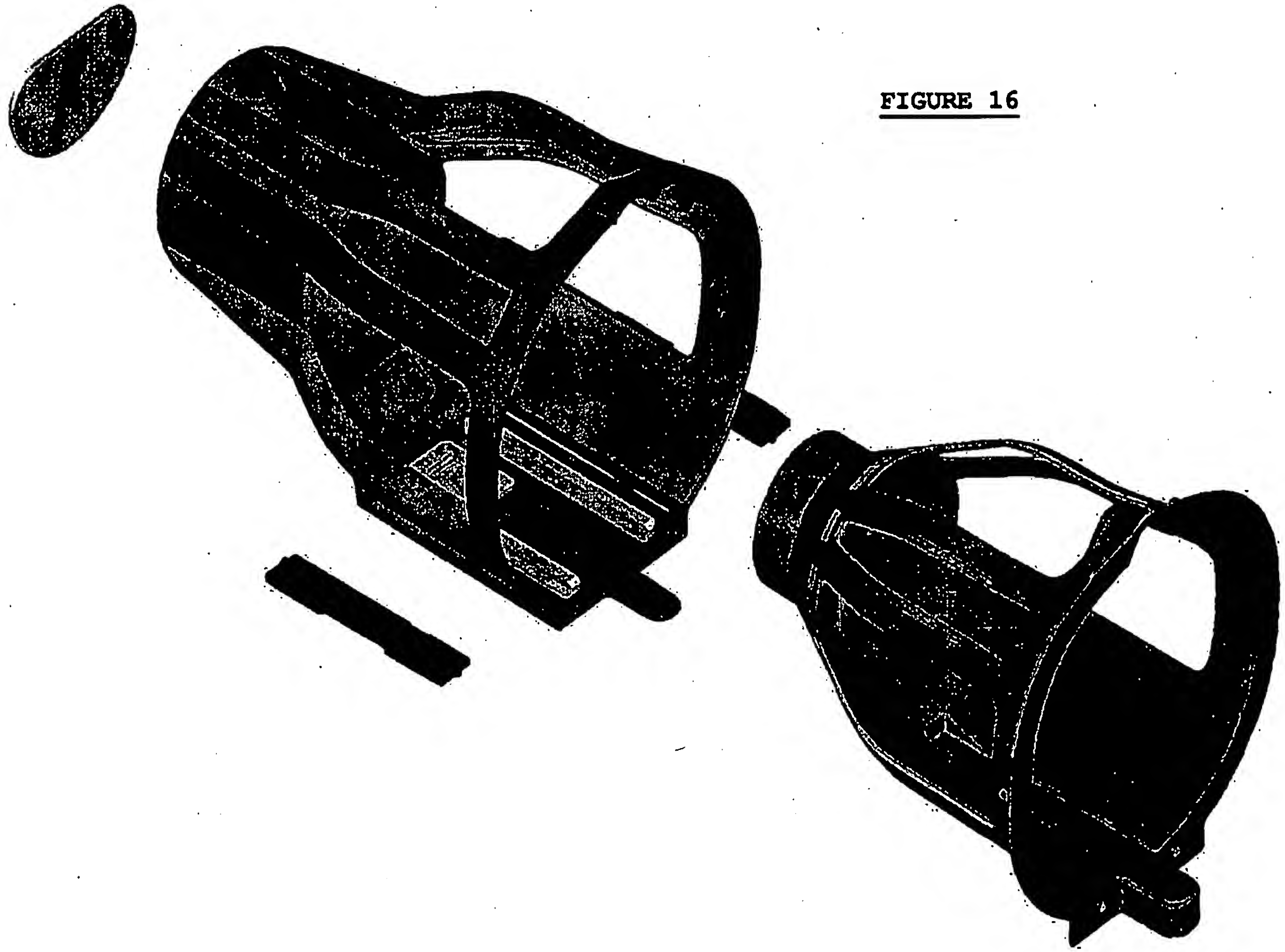


FIGURE 17

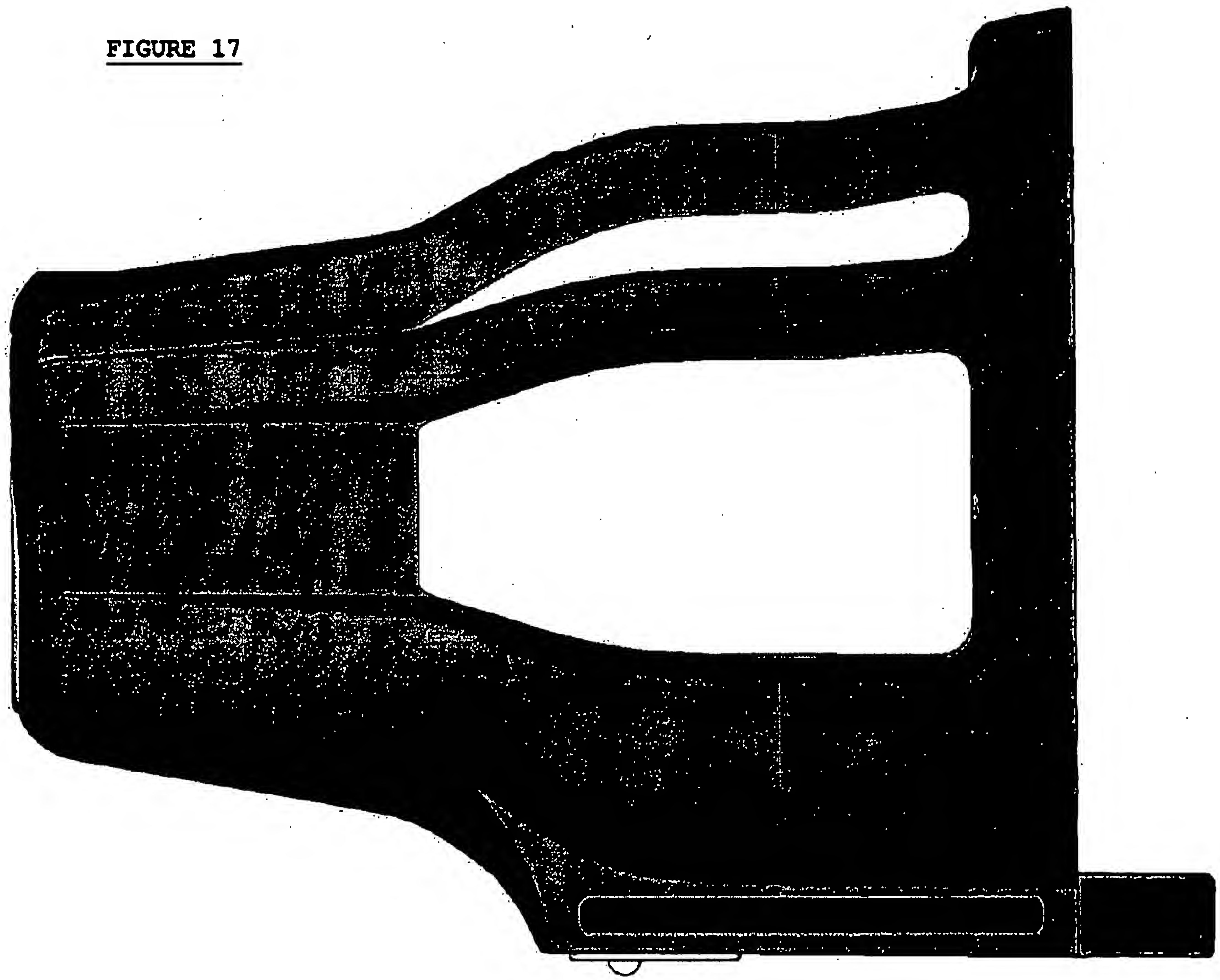


FIGURE 18

FIGURE 19

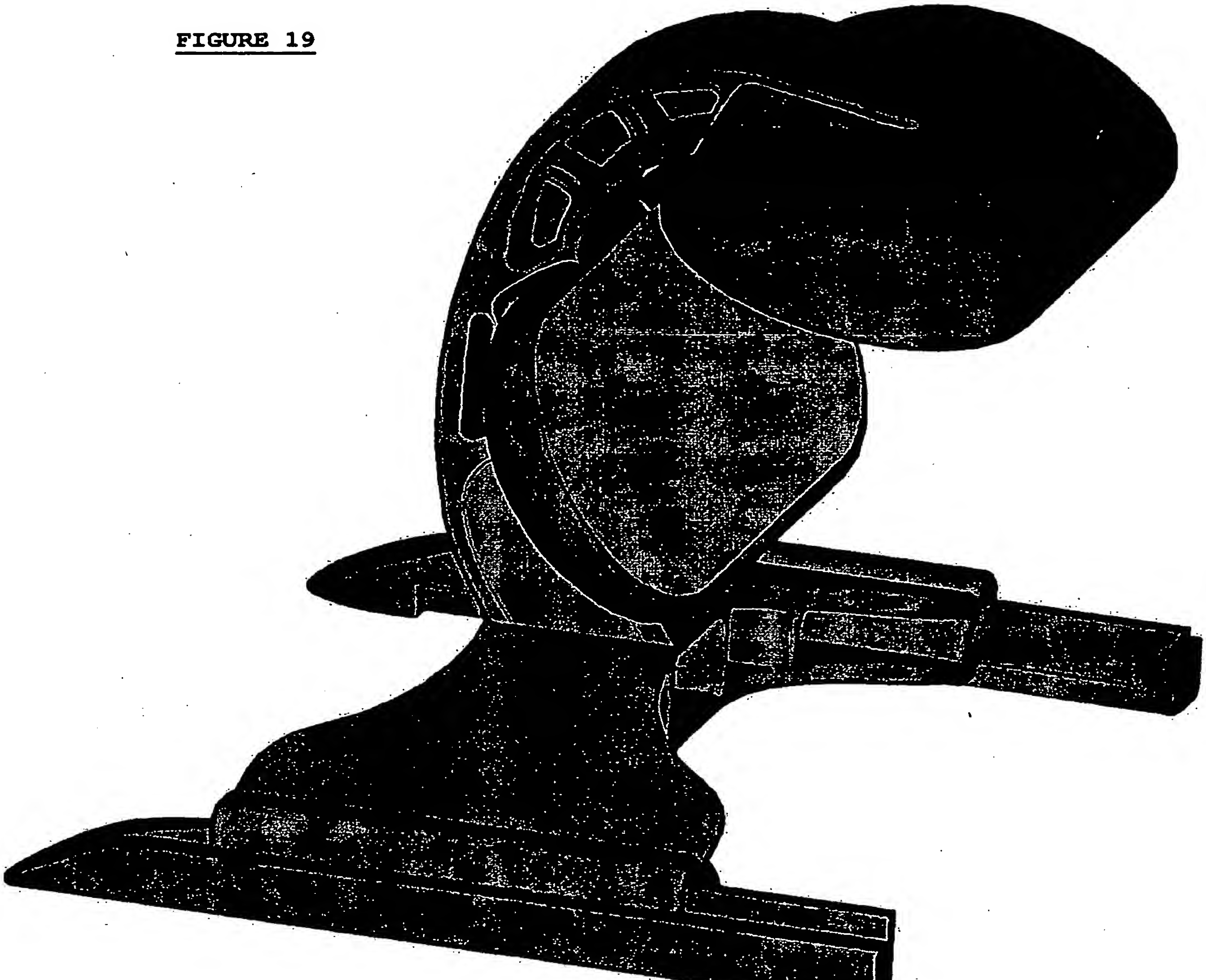


FIGURE 20

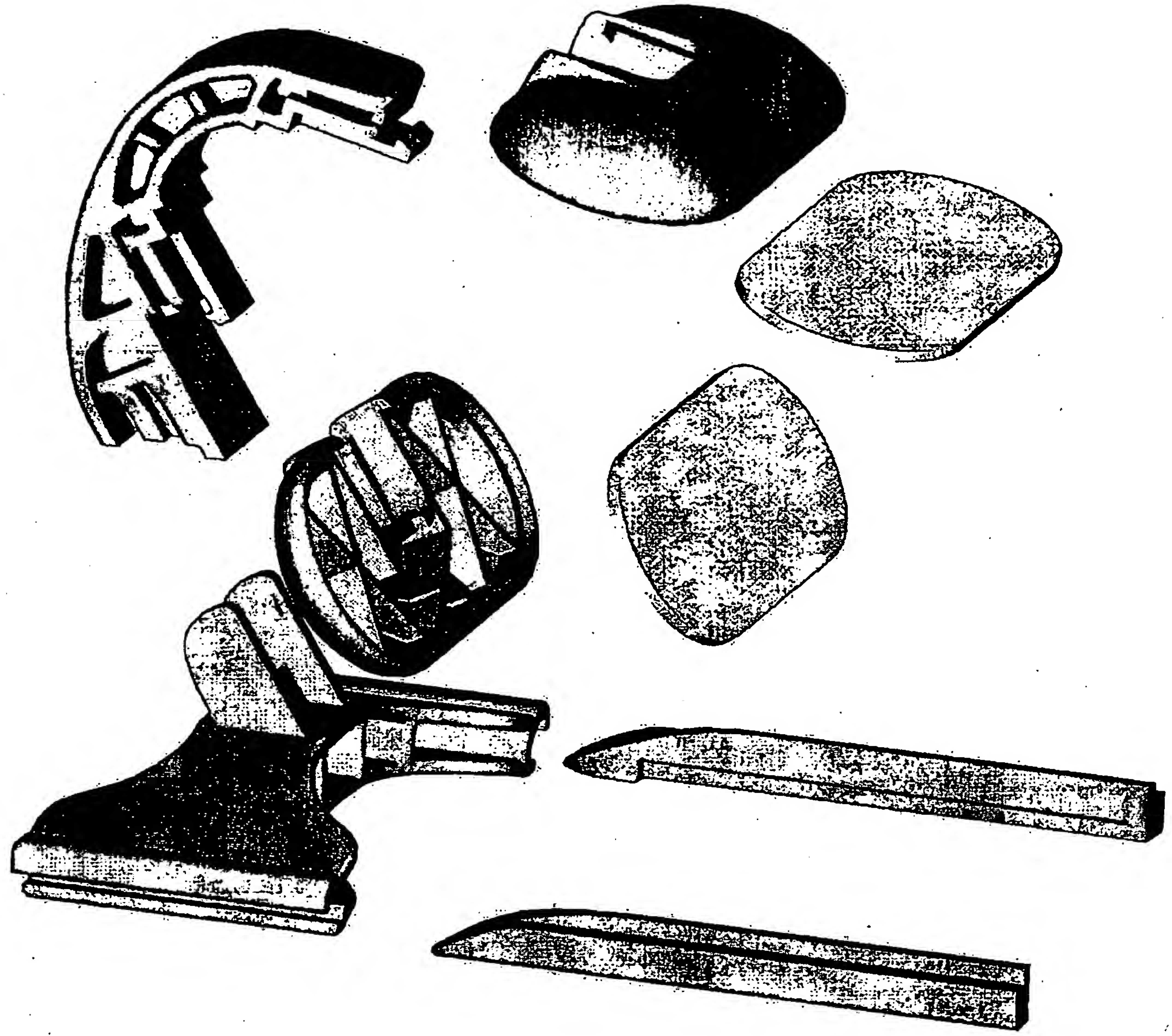


FIGURE 21

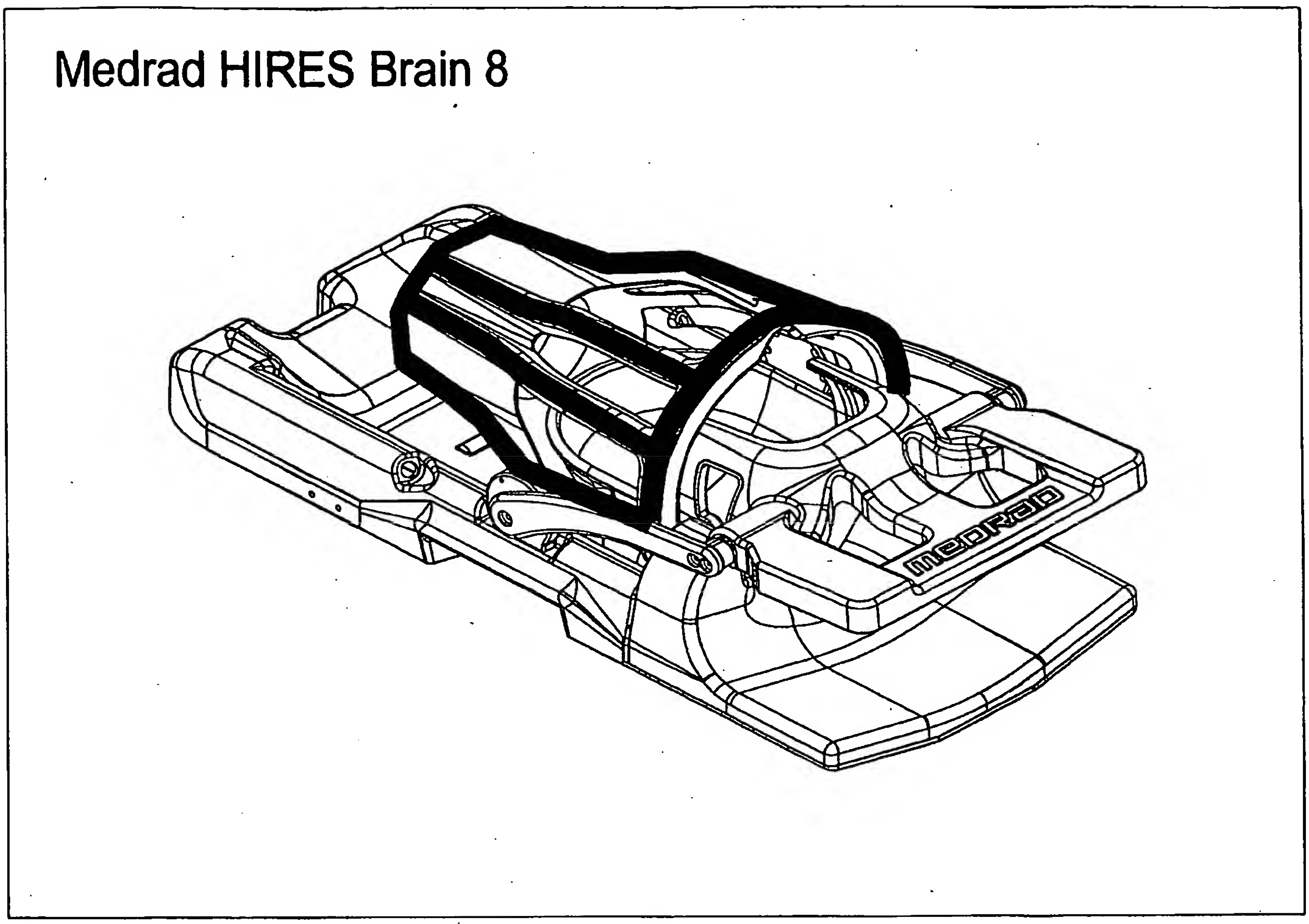


FIGURE 22

Medrad Anterior Neck

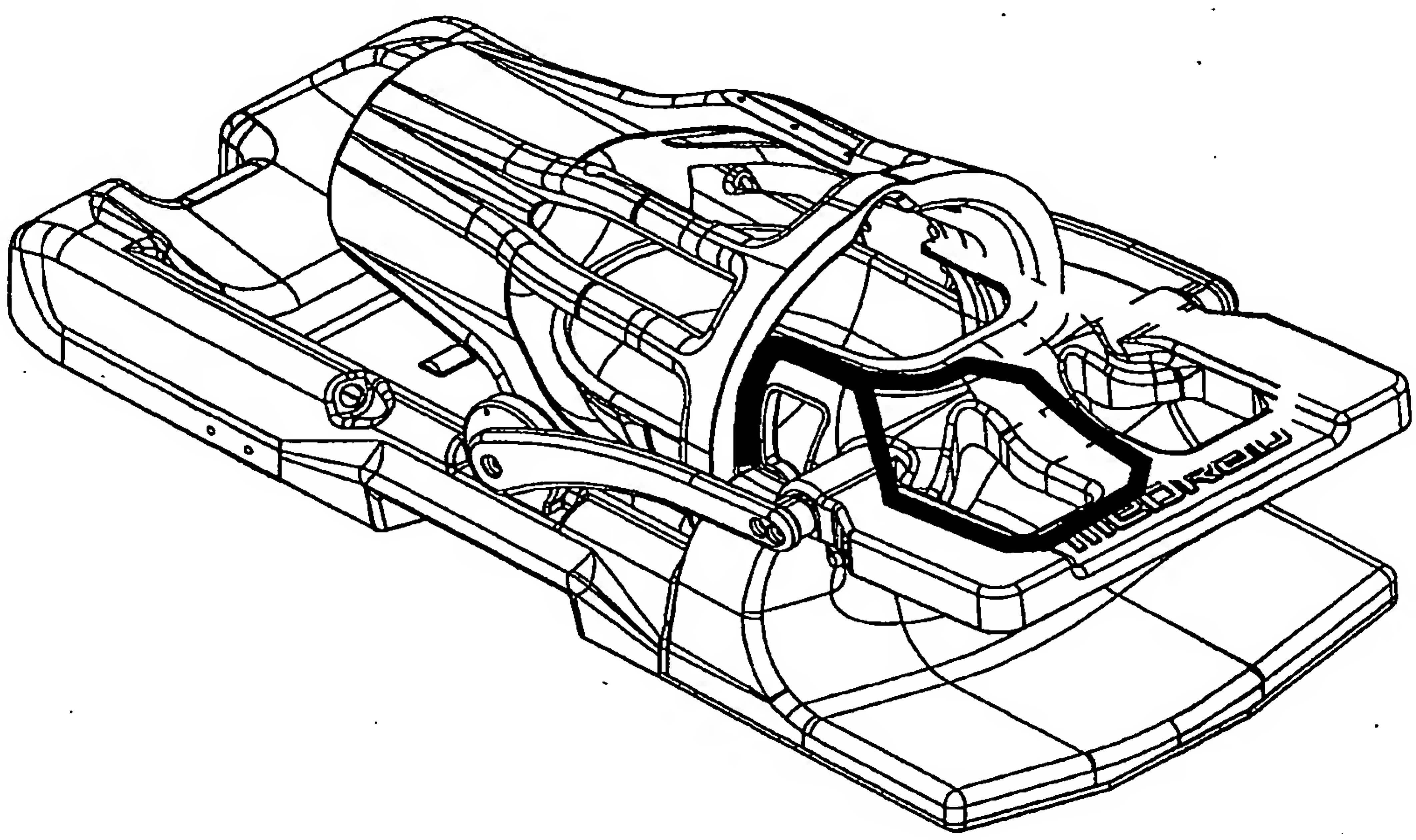


FIGURE 23

Medrad C-Spine

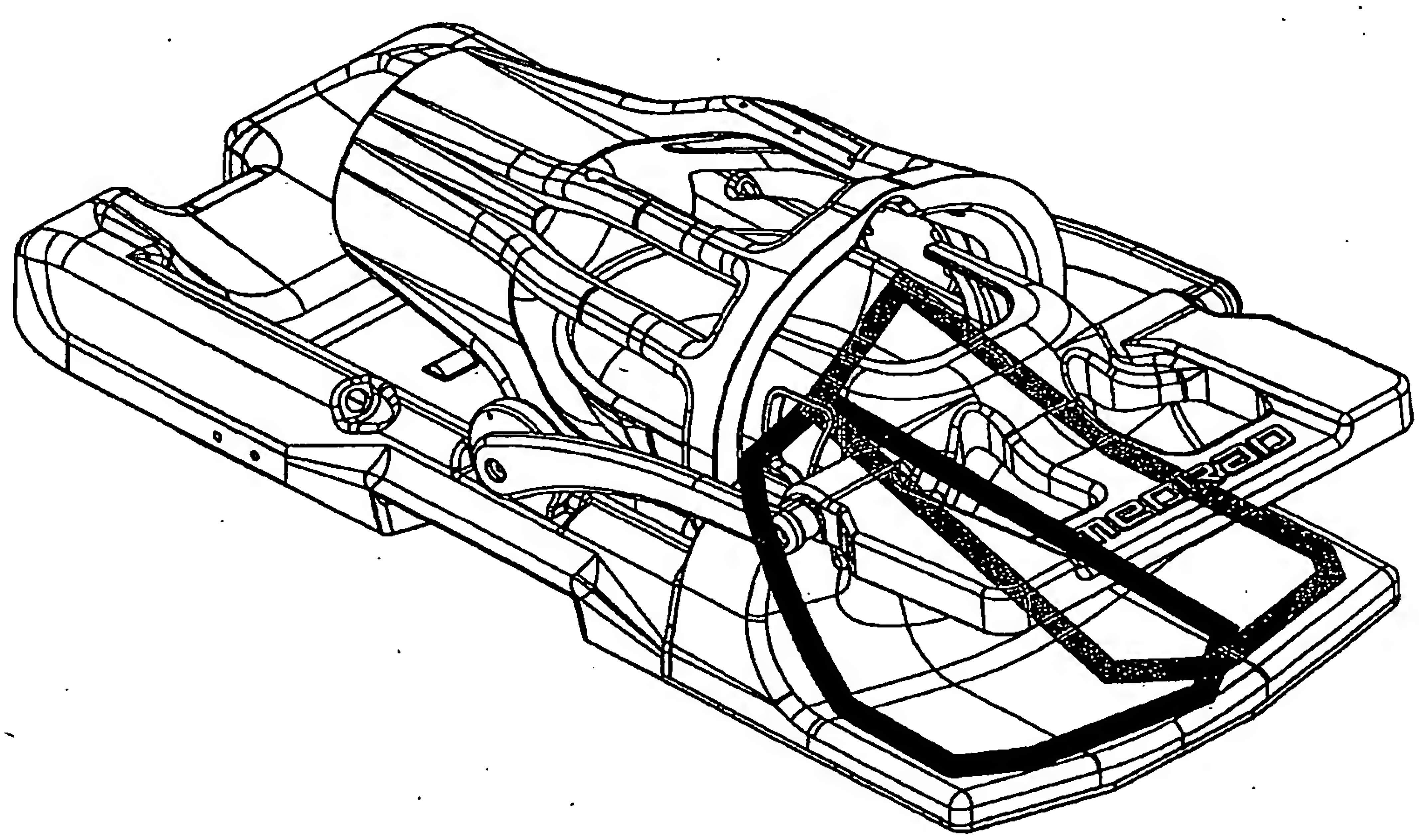


FIGURE 24

Medrad Volume Neck

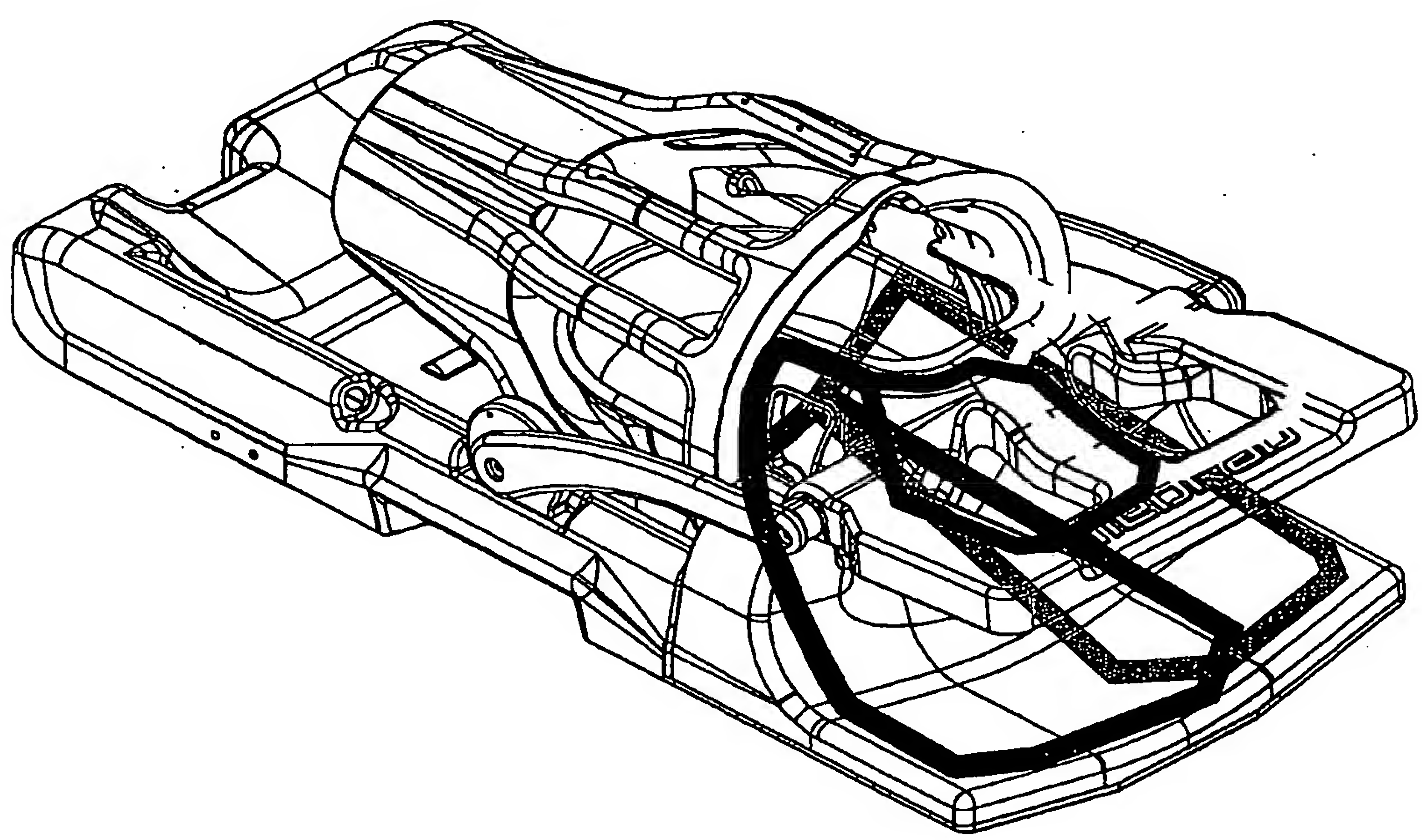


FIGURE 25

Medrad NVPA 8